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# A METHOD TO DEVELOP NECK INJURY CRITERIA TO AID DESIGN AND TEST OF ESCAPE SYSTEMS INCORPORATING HELMET MOUNTED DISPLAYS

DISSERTATION

Jeffrey C. Parr, Lieutenant Colonel, USAF

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**DEPARTMENT OF THE AIR FORCE**

**AIR UNIVERSITY**

## AIR FORCE INSTITUTE OF TECHNOLOGY

**Wright-Patterson Air Force Base, Ohio**

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DISSERTATION

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## Abstract

Helmet mounted displays (HMDs) are becoming common human-machine interface equipment in manned military flight, but introducing this equipment into the overall aircraft escape system poses new and significant system design, development, and test concerns. Although HMDs add capabilities, which improve operator performance, the increased capability is often accompanied by increased head supported mass. The increased mass can amplify the risk of pilot neck injury during ejection when compared to lighter legacy helmets. Currently no adequate US Air Force neck injury criteria exist to effectively guide the requirements, design, and test of escape systems for pilots with HMDs. This research effort presents a novel method to develop neck injury criteria to aid the design and test of future HMD-centric escape systems. The state of the art pilot-scale injury criteria risk functions developed in this research are constructed with combined human subject and post mortem human subject experimental data using a parametric survival analysis. The resulting neck injury criteria permit injury risk and classification levels specified by the Air Force escape system oversight office to be translated into system level test criteria. The application of the system level criteria during developmental and qualification testing of escape systems will ensure pilot safety and limit risk of neck injury. A Human Systems Integration analysis of the HMD trade space is also performed to demonstrate the importance of neck injury criteria and other tools to quantify the human-centric costs and benefits during HMD development.

**Acknowledgments**

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Jeffrey C. Parr

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## List of Abbreviations

| | | |
|---|---|---|
| AFB | …………………… | Air Force Base |
| AFLCMC | …………………… | Air Force Life Cycle Management Center |
| AFRL | …………………… | Air Force Research Laboratory |
| AIS | …………………… | Abbreviated Injury Scale |
| ATD | …………………… | Anthropomorphic test device |
| CG | …………………… | Center of gravity |
| DoD | …………………… | Department of Defense |
| DoDAF | …………………… | DoD Architectural Framework |
| ES-2 | …………………… | Eurosid-2 (ATD) |
| FAA | …………………… | Federal Aviation Administration |
| HMD | …………………… | Helmet mounted display |
| HSI | …………………… | Human systems integration |
| IARV | …………………… | Injury assessment reference value |
| INCOSE | …………………… | International Counsel on Systems Engineering |
| JSF | …………………… | Joint Strike Fighter |
| KEAS | …………………… | Knots equivalent airspeed |
| LR | …………………… | Logistic regression |
| MANIC | …………………… | Multi-axial neck injury criteria |
| ms | …………………… | Millisecond |
| NHTSA | …………………… | National Highway Transportation Safety Administration |
| NIC | …………………… | Neck injury criteria |
| NVG | …………………… | Night vision goggles |
| OC | …………………… | Occipital condyles |
| PMHS | …………………… | Post mortem human subject |
| SA | …………………… | Survival analysis |
| SE | …………………… | Systems engineering |
| SME | …………………… | Subject matter expert |
| TOC | …………………… | Total ownership cost |
| USAF | …………………… | United States Air Force |
| USN | …………………… | United States Navy |
| VDT | …………………… | Vertical Deceleration Tower |

A METHOD TO DEVELOP NECK INJURY CRITERIA TO AID DESIGN AND TEST OF ESCAPE SYSTEMS INCORPORATING HELMET MOUNTED DISPLAYS

# I. Introduction

## Overview

Helmet Mounted Displays (HMDs) are becoming common human-machine interface equipment in manned flight. They have been designed to increase the performance of operators in their weapon system and thus increase overall mission effectiveness (Rash et al., 2009; Booher, 2003). HMDs are currently in use on multiple Department of Defense (DoD) weapon systems; including the F-15, F-16, A-10, and F-18, as well as many rotary winged aircraft. These displays add capabilities such as enhanced night vision, faster data processing, and information fusion, all of which have the potential to enhance mission accomplishment across the spectrum of military operations.

The role of the HMD in the Air Force's (AF's) next generation fighter, the F-35, has been expanded. The HMD in the F-35 not only augments traditional in-cockpit displays and the Heads-Up Display, but replaces them with virtual instruments displayed only through the HMD. This design decision has a number of implications for the weapons system, the HMD, and the human operator. This virtualization of the avionics displays has the potential to simplify the cockpit design, improve maneuverability by removing weight from the front of the aircraft, and provide omnipresent avionics information to the pilot. Further, this HMD provides sensor information to the pilot's eyes permitting the pilot to view space around their aircraft, which would traditionally have been occluded by the airframe. While this feature has existed in fixed-wing aircraft, this change has the ability to provide the pilot with significantly improved situation awareness, particularly in missions such as close air support where visibility through the airframe constrained the angle of attack.

Unfortunately, the decision to integrate additional pilot information systems into the HMD leverages additional requirements on the display system, potentially increasing operator head supported mass compared to the mass of legacy flight helmets. Further, the HMD becomes such an integral part of the weapons system that the display cannot be removed from the helmet to remove mass as was possible with some legacy systems. This increase in helmet mass and persistence has the potential to degrade operator performance, health, and safety through increasing fatigue, increasing the potential for chronic neck injuries, and increasing the potential for operator injury during high acceleration events, such as those occurring during aircraft ejection. As such, virtualization of the cockpit display suite, first envisioned to improve overall system effectiveness, has the potential to induce initially unforeseen consequences that could potentially degrade the overall performance and safety of the system.

To ensure pilot safety is preserved during ejection, important parameters must be carefully implemented into HMD design. Increases in head supported mass can increase the forces placed on the operators' neck when exposed to accelerative environments. Therefore, the risk of operator neck injury can increase with increasing helmet mass as the pilot is exposed to highly accelerative environments, especially those that can be experienced during ejection. Studies performed with human subjects in accelerative environments have repeatedly demonstrated significant increases in neck loads when the subjects wear an HMD as compared to when the subjects do no wear one when exposed to the same input acceleration pulse (Buhrman and Perry, 1994; Perry, 1998; Doczy et al., 2004). Injury due to a heavier HMD with an off-axis center of gravity (CG) in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures and ligament ruptures (Buhrman and Perry, 1994). Perhaps this finding appears intuitive as increasing the mass of the head would be expected to result in an increased force when the head is exposed to acceleration. However, the human body

is a complex mechanical system including a series of linkages and soft tissue connections, which have the potential to dampen or amplify an input impulse.

Besides head supported mass, other HMD design parameters affect pilot neck loading and biomechanics including center of gravity (CG) and moment of inertia. Minimizing the weight of the HMD and distributing the mass of the components of the HMD such that the center of gravity remains as close to that of the head have been suggested as methods to reduce the likelihood and severity of pilot neck injury (Melzer, 2001). However, this requirement is often in conflict with the requirements to provide increasing capabilities in future HMDs, which often require placement of electro-optic components near the pilots' eyes, which are well forward of the center of gravity of the human head.

Pilot anthropometric factors may also affect the likelihood of injury from neck loads induced by head supported mass when exposed to acceleration; and recent changes in DoD manning requirements have increased the diversity of these anthropometric characteristics among pilots through the inclusion of non-traditional demographics (Harris, 1997). Systems must now accommodate pilots ranging from 103 lbs to 246 lbs (Nichols, 2006). Pilots at the lower end of this weight spectrum are usually females who have been documented to be more at risk of neck injury from neck loads induced by HMDs in accelerative environments because of their smaller neck bone structures and supporting musculature (Buhrman and Wilson, 2003; Perry, 1998). Smaller pilots are required to support a proportionally higher HMD mass when compared with their overall body mass. Take for example the following HMD weights from an existing DoD HMD design: small HMD - 4.49 lbs, med HMD - 4.56 lbs, and large HMD - 4.64 lbs. The ratio of body mass to helmet mass can be calculated using the average masses of the standard representative human, which are 115.5 lbs for small, 167.5 lbs for medium, and 222.5 lbs for large pilots (Nichols, 2006). Ratios of HMD mass to body mass would then be 3.89%, 2.72%, and 2.09%, respectively. The ratio of HMD mass to body mass for the lowest weight

individuals is nearly twice the same ratio for larger male pilots. Therefore, it is important that pilot neck response due to heavier HMDs be understood and characterized using a standard evaluation criterion while understanding the influence of pilot anthropometric and biomechanical characteristics on the likelihood of injury.

### Framing the Problem within Systems Engineering and Human Systems Integration

It is important to frame this topic area within the context of the field of systems engineering (SE) as a whole. Within the field of systems engineering, Human Systems Integration (HSI) represents an overarching methodology to ensure systems are designed and built to maximize performance and minimize total ownership cost (TOC) by considering the human as a critical element of the system. The International Council on Systems Engineering (INCOSE) Handbook defines HSI as “the interdisciplinary technical and management processes for integrating human considerations within and across all system elements; an essential enabler to systems engineering practice (INCOSE, 2011).” The domains of HSI using the structure put forth by Miller et al. are depicted in **Figure 1**.

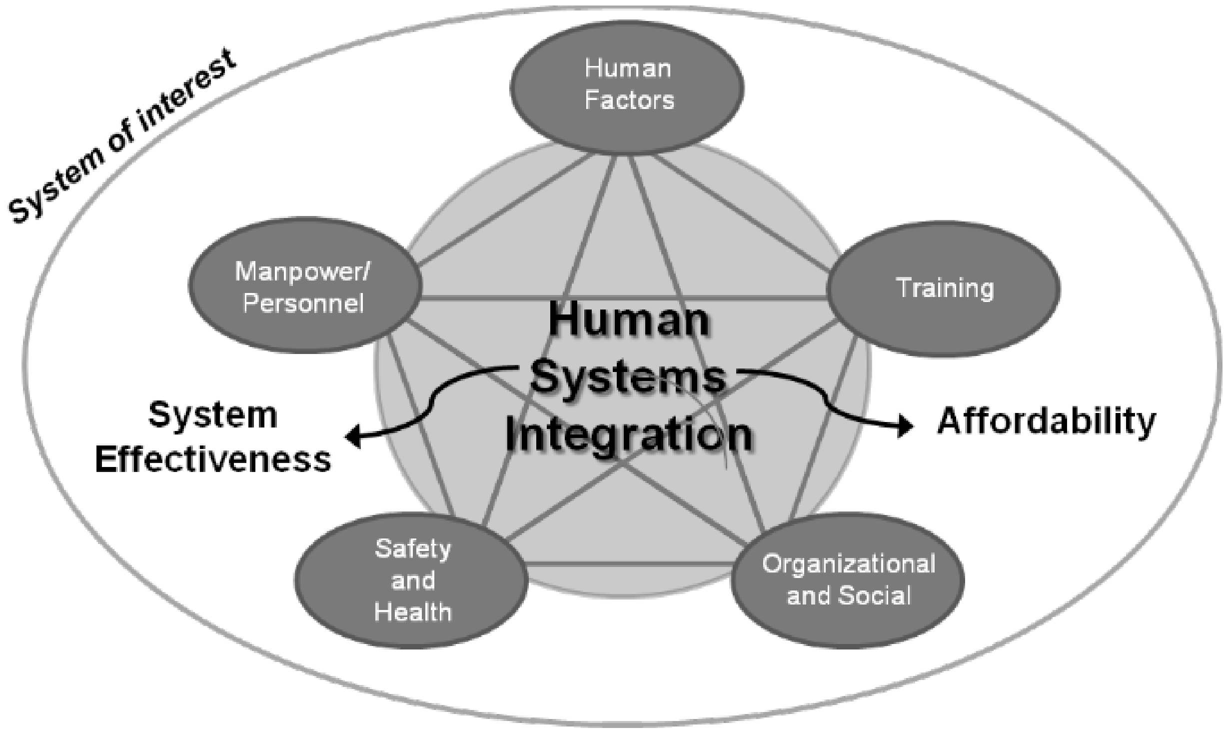


**Figure 1. The Goal of HSI is to Integrate the Domains (from Miller et al., 2013)**

The benefits of addressing the HSI domains and incorporating them into system design early in the systems acquisition lifecycle are numerous and well documented (Booher, 2003; Hardman, 2009). The traditional SE "V" (**Figure 2**) depicts the SE processes from user/operation requirements at the beginning of the acquisition process to transition. Early implementation of HSI into the SE processes as shown in the V model is expected to "result in increased weapon system safety, reduced life cycle costs, and optimized weapon system performance (Air Force Human Systems Integration Office, 2009)."

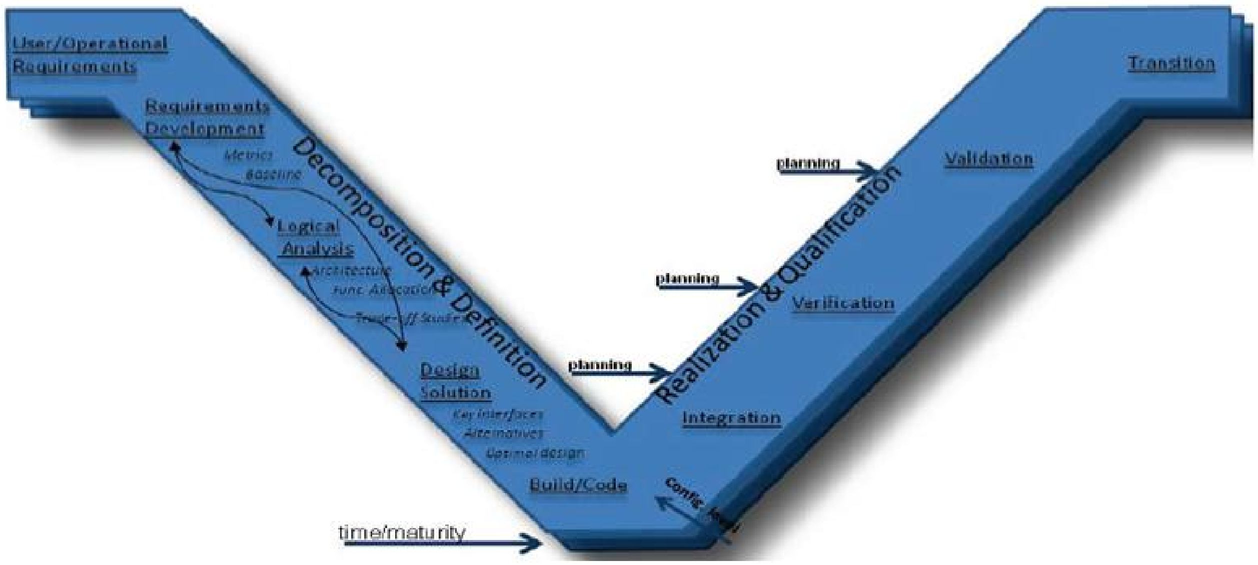


**Figure 2. SE Processes on the V Model (from Hardman, 2009)**

The Management Guide to HSI in Acquisition states "The goal of HSI is to maximize total system performance, understanding that the human element is an integral part of systems, while minimizing total ownership costs. To be effective, HSI must be conducted as a fundamental part of the overall systems engineering activities within the Air Force Integrated Defense Acquisition, Technology, and Logistics Life Cycle Management System (HSI in Acquisition, 2009)." The Human Factors domain of HSI encompasses "the comprehensive integration of human capabilities…into system[s] (Air Force Human Systems Integration Office, 2009)." Most physical aspects of HSI fall into the Human Factors domain, and the biomechanical topics pursued in this work exist in this arena. The pursuit of neck injury criteria and the desire to protect pilots from neck injury due to the added weight of HMDs in accelerative environments fall under the Safety and Health domain. Knowledge gained by research on human neck response to HMD systems can help the DoD better design, develop, modify, and evaluate HMD and aircraft escape systems to optimize human performance and ensure pilot safety. Neck injury criteria are applied at various stages in the acquisition process as shown in **Figure 3**, specifically during the technology development phase and the engineering and manufacturing development phase.

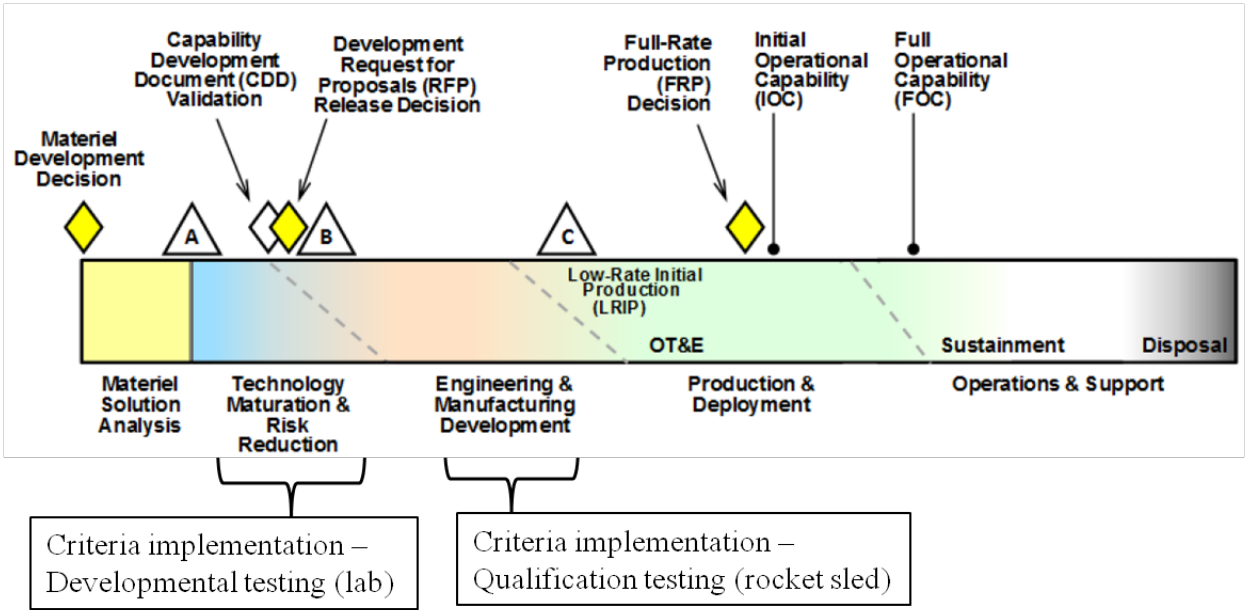


**Figure 3. Implementation of Neck Injury Criteria within the DoD Acquisition Timeline (DAU, 2013b)**

It is important to point out that HSI is often not performed well because, unlike trades in hardware/software systems, where good physical and behavioral models exist, robust physical and behavioral models often do not exist for the human element. Therefore, these trades are conducted by either applying logical processes, which often fail to identify significant risks or through iterative prototyping and testing, which can be expensive and that require robust test methods which take considerable time to develop. This difficulty leads into the primary motivation for the research performed in this dissertation.

### Motivation

The primary motivation behind this dissertation research is the lack of comprehensive, multi-axial, aviation-specific neck injury criteria for accelerative environments that is satisfactory to the acquisition escape community requirements. Without comprehensive criteria based upon well constructed risk functions, HMD and escape system designers are left without design guidance to produce HMDs and escape systems with appropriate mass properties that are

safe for use by pilots. System capability is likely to be suboptimal without a method to quantify the safety risk associated with these systems.

## Research Goals

This research aims to provide AF acquisition personnel with improved risk criteria to help develop safe HMDs and escape systems. Specifically, this research develops and proposes a method for the construction and development of improved aviation-specific, ejection neck injury criteria as well as applicable metrics for these criteria. Additionally, this research places these metrics within the larger context of HSI by incorporating these criteria components within a general HSI model of HMD capability versus added mass to attempt to quantify the safety impacts of HMD mass on the system. Injury risk curves serve as the basis for any neck injury criterion (Pellettiere, 2012). Thus developing improved neck injury risk functions are a major component of this research. The research objectives are as follows:

1) Present a method for the development of improved aviation-specific ejection neck injury criteria as well as constructing feasible pilot study criteria that can be used to evaluate and mitigate risk posed by various HMD and escape system configurations.
    a. Develop the Criteria's Underlying Fundamentals: Develop the structure and components of the neck load input variables. Determine appropriate statistical methods. Determine data requirements.
    b. Create the Risk Functions: Develop human risk functions which help understand the risk to an individual in an environment where they are exposed to high acceleration in all possible axes.
    c. Apply the Risk Functions: Compare performance of new criteria to legacy criteria within a set of ejection system testing data.

2) Provide decision makers with an improved method to conduct HSI safety trade studies during the HMD acquisition process.
    a. Apply the risk functions to quantify the safety impacts of various HMD masses using existing data.
    b. Perform preliminary qualitative HSI trade analysis of HMD capability versus safety.

The first objective of the method involves establishing the supporting structure and fundamentals required for the development of improved neck injury criteria and construction of the risk functions that serve as the foundation of the criteria. This includes developing the appropriate structure of the neck load inputs based upon the current state of the art in neck injury research, applying the appropriate statistical methods to the data for optimum risk function development, and identifying the data required to generate robust risk functions from the appropriate sources (human subject and PMHS). Also necessary to accomplish the first objective is applying the structure and fundamentals to the development of risk functions.

**Current Practices**

Air Force personnel involved in HMD acquisition, research, and development are currently employing limited criteria during system design. One criterion used to guide prototype HMD design is the Knox Box. The Knox Box was provided by the USAF ejection research community to the acquisition community nearly 20 years ago. This criterion was originally intended to serve as an interim criterion for HMD weight and center of gravity. It provides guidance on specific design-related attributes of an HMD (e.g., mass and CG) given that a host of assumption are met. However, its use has persisted as it is the only criterion that is applicable early during the design stage. Once an HMD prototype is designed, developed, and technically mature, other evaluation criteria are implemented. The entire escape system is evaluated in a

series of sled tests with anthropomorphic test devices (ATDs) recording forces in simulated ejections at flight speeds on a rocket sled at Holloman AFB. These evaluation criteria are applied to assess the safety performance of the escape system, which includes the HMD as part of the pilot's ensemble.

The most current example of this developmental testing process is from the F-35 program where the entire system is new and the seat, HMD, canopy, and all other components of the escape system must be tested to pass safety requirements. The US Navy (USN), as the lead agency for the F-35 development, applied a 12-part neck injury criterion, hereafter referred to as the USN Neck Injury Criterion or NIC (Nichols, 2006). Instrumented aerospace ATDs fitted with Hybrid III necks (a standard neck load cell configuration used in both automotive and military testing) representative of large, medium, and small pilots undergo the full ejection sequence launched from a rocket sled at variable knots equivalent air speeds (KEAS). The measured ATD neck loads are compared to the NIC limits, and the system is accepted or failed based upon this data. It should be noted that cross-validation of the Knox Box with the NIC has not been formally provided. Anecdotal evidence from observers show that no HMD designed using the Knox Box has failed developmental rocket sled testing (Mattis, R. Personal Communication; 2013). On the other hand, HMD systems outside of the Knox Box have passed system level developmental rocket sled testing. Therefore, it is possible that designs which are outside the Knox Box might or might not pass the NIC, which can then lead to a costly iterative design process as components are rejected during developmental test after the system has been designed and constructed.

Additionally, the NIC has some flaws that have raised questions as to how well it is able to evaluate systems from a design and risk assessment standpoint. The first flaw is that it can provide internally conflicting load limits. As a result, a system may pass some criteria and not others, at which point the acceptability of the system is ambiguous. This fact points to the

serious problems posed by the criteria to HMD designers. Inconsistent criteria impede and confuse the design process. Another flaw is that the NIC incorporates sub-criteria elements that have no associated risk functions tied to specific injury levels. This makes it unclear whether a system that does not meet the suggested limit will result in an unacceptable level of injury. Additionally, the NIC contains elements that allow for greater risk of injury than the AF escape system oversight office allows. Each of these issues complicates the system design process and presents an opportunity to develop improved Air Force-specific neck injury criteria that could help support the improvement of the escape system developmental testing standard.

The process as it stands now affords a very limited HMD design window based upon known safe loads for human subjects. Additionally, the Knox Box provides no neck injury risk information, only a maximum HMD mass specific to ejection seat type and a small range for deviation of the combined center of gravity of the HMD and head. Very little other HMD design guidance is available to HMD and escape system program managers and defense contractors to guide safe design of HMD mass properties in the context of the overall escape system.

The escape system oversight office of the Air Force Life Cycle Management Center (AFLCMC), which is the USAF aircraft acquisition center, has clarified neck injury criteria requirements for USAF aviation. They have specified that multi-axial neck injury criteria be developed to evaluate HMDs and new escape systems such that acceptable injury rates should be 5% at an injury classification level of Abbreviated Injury Scale (AIS) 2 (moderate injury) compared to the 10% risk at AIS 3, which is loosely incorporated into some sub-elements of the NIC (Parr et al., 2013). They have additionally requested that the improved criteria be tied to clearly defined probability of injury, which they assert are very important to decision makers in the systems engineering and acquisition process. No existing neck injury criteria meet these requirements, and this work will attempt to provide a method for addressing this need of the USAF acquisition community.

## Research Implications

If improved, aviation-specific, ejection neck injury criteria are developed and validated, it is possible that significant savings could be realized in the HMD acquisition process with standard and consistent neck injury criteria in place to which to design. This will also provide important risk functions that will provide insight into the effect of HMD mass properties on pilot safety. The cost of developing prototypes of HMDs and other escape systems that could possibly fail the currently conflicting and redundant neck injury criteria could be avoided. The current NIC is expensive to evaluate. An exceedence of any of the criteria triggers a subject matter expert (SME) review, which is a cost as well as a schedule hindrance. Also, the contractor does not immediately know the results post test since it may or may not be a failure; this makes it difficult to make decisions on how to proceed. Besides cost, it is also possible that other benefits may be realized. The current approach significantly limits the trade space for HMD systems. The presence of enhanced, aviation specific neck injury risk criteria could potentially expand the design space, permitting the inclusion of more capable head mounted systems in future cockpits.

Primarily, users, program managers, and engineers involved in the operation and acquisition of HMDs are concerned with designing and fielding systems that are safe but also provide the most capability technology will allow. The AFLCMC escape systems oversight office who requested the development of an improved criterion and escape engineers who test and qualify these systems are also key stakeholders in this research effort. This work may also influence non-DoD crash safety system design (e.g., automotive, off road vehicle, and sports).

## Assumptions/Limitations

This work will use existing human and PMHS neck load and injury data to create pilot-scale risk functions and criteria, to understand the limitations of the current data, and to demonstrate a method for development of robust functions and criteria. Final form neck injury

criteria designed to replace the existing escape system developmental testing criteria (the NIC) will require future research to build upon the foundation this research provides. References are made in this work to qualification testing criteria since this is the final goal of this line of research and improving these criteria is the motivation for this work. The pilot-scale axis specific injury criteria (Gx, Gy, Gz) developed to meet the AFLCMC escape office's requirements in this work will be able to provide injury risk information specific to those axes of acceleration and can be used to evaluate systems in the single axis, as well as provide risk prediction based upon neck loads from variable head supported masses. It is anticipated that this pilot-scale criteria will provide a foundation to advance toward a final AF neck injury criterion for use in qualification testing. In this research, when the criterion is applied to test data observed using an ATD, it will be assumed that the ATD and human neck responses are similar for preliminary assessment of the criterion, though it is known that some researchers have observed instances where the Hybrid III ATD neck response is not completely biofidelic. Future work outside the scope of this dissertation would be required to make the pilot-scale criterion developed in this work fully applicable to a system evaluation of a complete ejection sequence using a surrogate. This future work would include the development of human to Hybrid III ATD neck transfer functions and further experimentation with ATDs to validate the transfer functions.

**Research Questions and Hypothesis**

This research addresses the following research questions: What is an appropriate structure and formulation for improved aviation specific neck injury criteria, and what data exists or is required to adequately establish these criteria? What methods should be followed to develop adequately supported multi-axial aviation specific neck injury criteria? The research hypothesis is "It is possible to develop aviation-specific ejection neck injury criteria and a human tolerance validated metric for the criteria that can be used to evaluate and mitigate risk posed by

various HMD configurations and escape system development as well as provide decision makers with information to conduct safety trade studies during the HMD and escape system acquisition process."

### Dissertation Structure

A modified scholarly approach is employed in this dissertation. Much of the content is comprised of papers that have been submitted or accepted for publication in peer reviewed conference proceedings or journals. These papers are located within the dissertation to document the research process, either as dedicated chapters or inserted into suitable sections of a chapter. Where necessary, additional content is interspersed to frame the papers and integrate them into the overall dissertation. **Table 1** provides a summary of the scholarly publications related to this research.

**Table 1. Summary of Scholarly Publications**

| Paper Title | Forum | Status | Authors | Location |
|---|---|---|---|---|
| Evaluation of the $N_{ij}$ Neck Injury Criteria with Human Response Data for Use in Future Research on Helmet Mounted Display Mass Properties | Human Factors and Ergonomics Society 56th Annual Meeting, 2012 | Published in conference proceedings | Parr et al. | Chapter II |
| Neck Injury Criteria Formulation and Injury Risk Curves for the Ejection Environment: A Pilot Study | Journal of Aviation, Space, and Environmental Medicine | Published in journal Dec 2013 | Parr, Miller, Pellettiere, Erich | Chapter IV |
| Development of a Side Impact (Gy) Neck Injury Criterion for use in Ejection System Safety Evaluation | IIE Transactions on Occupational Ergonomics and Human Factors Journal | To be submitted for journal publication | Parr, Miller, Colombi, Schubert-Kabban, Pellettiere, | Chapter V |
| Development of an Updated Tensile Neck Injury Criterion | Journal of Aviation, Space, and Environmental Medicine | Accepted for journal publication | Parr, Miller, Schubert-Kabban, Pellettiere, Perry | Chapter VI |
| Neck Injury Criteria to Aid Design and Test of Helmet Mounted Display Systems | Journal of Biomechanical Engineering | To be submitted for journal publication | Parr, Miller, Colombi, Schubert-Kabban, Pellettiere | Chapter VII |
| A Human Systems Integration Analysis of Helmet Mounted Displays | SAFE Journal 2014 | Accepted for journal publication | Parr, Miller, Colombi | Chapter IX |
| A Human Systems Integration Analysis of Helmet Mounted Displays | SAFE Conference 2013 | Published in conference proceedings | Parr, Miller, Colombi | Appendix A |
| Assessment of the Applicability of the NHTSA $N_{ij}$ Neck Injury Criteria to the Ejection Environment | Aviation, Space, and Environmental Medicine Conference 2013 | Published in conference proceedings | Parr and Miller | Appendix B |

This chapter presented an overview of the dissertation. The second chapter of this dissertation provides pertinent background information on HMD mass properties, neck biomechanics, research conducted to date on neck loading and response to accelerative environments, risk curve development, and neck injury criteria. The third chapter presents the research methodology applied to develop risk functions for each of the three primary axes of acceleration (Gx, Gy, and Gz). Chapters IV, V, and VI apply the methods from Chapter III and present results and discussion in the form of papers that have been published or submitted to peer-reviewed conference proceedings or scholarly journals. Chapter IV develops the –Gx axis of acceleration risk function, Chapter V develops the Gy axis of acceleration risk function, Chapter VI develops the Gz axis of acceleration risk function in the form of a single force, tensile risk function, and Chapter VII presents the complete multi-axial neck injury criteria (MANIC), which is a combination of the Gx, Gy, and Gz sub-criteria. Chapter VIII incorporates the AF ejection neck injury criteria system and stakeholders into the DoD Architecture Framework. Chapter IX presents a HSI analysis of HMDs, outlining the trade space and proposing a preliminary model to maximize the ratio of total system performance and TOC. Chapter X provides conclusions and makes recommendations for future research.

## II. Background

*"The cervical spine is one of the most complex structures in the human skeleton and its behavior during impact is still poorly understood* (Meyer et al., 2004)."

The foundational research of neck injury thresholds, tolerance to impact, strengths and biomechanical properties of biological materials, and injury pathways were initially accomplished for use by the automotive research community (Mertz and Patrick, 1971; Yamada, 1973; Sances et al., 1981; Brinn et al., 1986). The military aviation community began designing and building ejection seats into high speed aircraft in full force after World War II. As pilot safety became increasingly important and additional head supported mass (helmets, oxygen masks, etc.) became common during and after the Vietnam War, neck injury risk mitigation took on greater importance to the Air Force operational and research communities. The automotive neck injury body of knowledge was adopted, applied, and expanded by research performed in highly accelerative, ejection-like environments at AFRL on ATDs, human subjects, and PMHS to understand the biomechanical effects and injury pathways and thresholds on the neck muscles, ligaments, tendons, and vertebrae. The incorporation of helmets and HMDs spawned research by the DoD into the impact additional head supported mass would have on pilot safety. Additionally, program managers of recent weapon systems acquisition programs have faced decision making challenges. Human Systems Integration tradeoffs between capability and safety must be made, and very little work has been performed to quantify capability and safety to aid the decision making process. Developing injury risk curves and neck injury criteria relevant and applicable to the aviation environment with head supported mass is an important step to quantify safety risks. Each of these topics will be addressed in this chapter.

**Head and Neck Anatomy**

First, it is important to understand human head and neck anatomy. The human head weighs on average between 9 and 10 lbs. Plaga and Albery summarized existing PMHS head mass property literature values to propose new ATD head mass properties (Plaga and Albery, 2003). Literature values for average adult head mass ranged from 7.27 to 9.83 lbs. Plaga and Albery concluded that specifications for ATD heads based upon human data should be 7.4 lbs for the 5th percentile female, 8.1 lbs for the 50% male, and 11.0 lbs for the 95th percentile male (Plaga and Albery, 2003). These head masses are used on current aerospace ATDs for escape system qualification testing (Nichols, 2006). The head rests on and is attached to the cervical spine (neck) at the occipital condyles (OCs). The occipital condyles also provide a point of rotation for the head about the neck and are a landmark used in the field of biomechanics from which to measure upper neck moments in human subjects or ATDs (Chancey et al., 2007). The CG of the head is forward of the OCs; the head is prevented from falling forward by a counter force provided by the dorsal neck muscles.

The cervical spine consists of four separate units that contribute to the overall function of the complete unit; they are 1) the atlas, 2) the axis, 3) the C2-3 junctions, and 4) the remaining C4-9 vertebrae (Bogduk and Mercer, 2000). The atlas is the cradle for the OC, is very strong, and allows only nodding movements. The axis bears the weight of the atlas and allows for added axial rotation. The C2-3 junction is also called the root of the cervical spine, and is known as the start of the cervical spine. It functions and looks like a deep root which anchors the structures above (atlas and axis) to the structure below (the remaining C4-9 vertebrae) (Bogduk and Mercer, 2000). The remaining C4-9 vertebrae are shaped and move alike, stacked similarly with intervertebral disks in between each vertebra. Each of the vertebrae that make up the complete cervical spine is connected by ligaments and fibrous capsules making up the soft tissue of the cervical spine (Yoganandan et al., 2001).

Adding mass to the head in the form of an HMD and/or altering the natural CG of the head (specifically moving the CG forward) can have negative effects on the natural kinematics of the human cervical spine (Bogduk and Mercer, 2000). Neck muscles, ligaments, and bones accustomed to supporting a fixed weight are called upon to provide support and stability to a new configuration, which subjects these structures to increased force and potential injury. Neck muscles involved in head stability and locomotion include sternocleidomastoids; longus colli and capitis; scalenus anterior, medius, and posterior; trapezius; semispinalis capitis and cervicis; longissimus capitis and cervicis; and the splenius capitis and cervicis (Teo et al., 2004). When humans are exposed to accelerative forces with head supported mass, effects on the neck range from fatigue, minor neck soreness, and pain to severe neck injury or death. The next section provides a summary of the literature related to the effects of head supported mass and HMDs on head and neck biomechanics in accelerative environments.

**Neck Biomechanics in Accelerative Environments**

When a pilot ejects from an aircraft, he or she is subjected to four different phases, each phase exposing the pilot's head and neck to different forces. In order, these phases are: catapult stroke, windblast, seat stabilization, and parachute opening. At all stages of the highly dynamic ejection sequence, the neck can be subjected to any of the primary forces which include axial loading [tension ($+Fz$) or compression ($-Fz$)], frontal shear ($Fx$), side shear ($Fy$), anterior/posterior bending [flexion ($+My$), extension ($-My$)], side bending ($Mx$), and twisting ($Mz$). During catapult stroke the primary forces acting on the neck are compression and flexion from the high +Gz acceleration (see **Figure 4** for anatomical coordinate system). Windblast exposes the pilot to large tensile forces (from high lifting forces on the head and helmet), while seat stabilization can potentially expose the pilot to flexion and tension (due to high –Gx acceleration) where primarily neck compression and flexion occur (Pellettiere et al., 2005).

Finally, parachute opening shock can subject the pilot to both Gx and Gz acceleration. Additionally, depending on the orientation of the aircraft at the time of ejection, substantial sideward (Gy) forces are also likely to be present throughout the ejection sequence. Most aviation-specific ejection studies have focused on the effects of the first phase, catapult stroke, in which the accelerative forces presented by the ejection mechanism act upon the head and neck in the positive z axis (upward, or +Gz). However, the accelerative forces during all phases of ejection are a concern as head supported mass has increased due to the introduction of helmet mounted equipment like night vision goggles (NVGs), advanced optics, and other HMD components to aircrew helmets. As a result, numerous studies have been conducted to evaluate ATD and human neck response to a range of head supported masses with various CGs (Buhrman and Perry, 1994; Perry, 1994; Perry and Buhrman, 1995; Perry and Buhrman, 1996; Perry et al., 1997; Perry, 1998; Buhrman and Wilson, 2003; Salzar et al., 2009). Other research has evaluated neck response from other phases of ejection, which include exposure to frontal (-Gx) and sideward (Gy) acceleration (Buhrman and Mosher, 1999; Perry et al., 2003; Doczy et al., 2004). The next section provides a summary of this research.

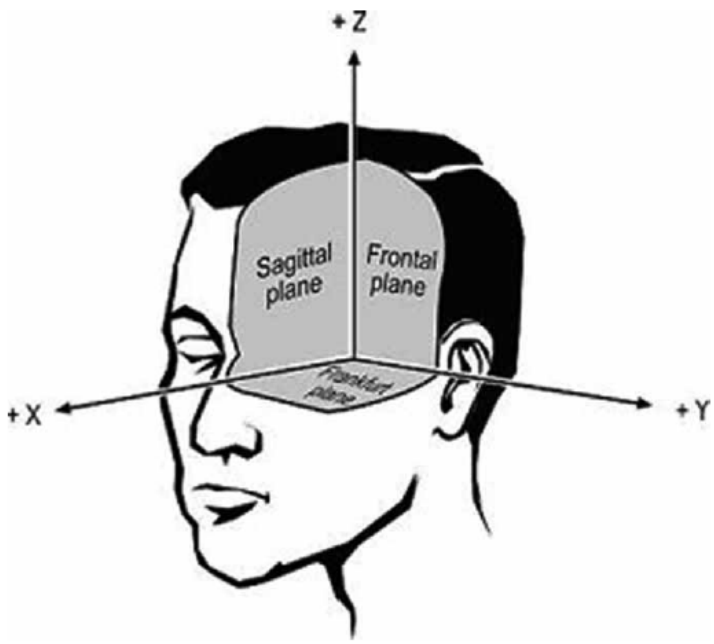


**Figure 4. Anatomical Coordinate System of the Head** (Rash et al., 2009)

**Non-injurious Accelerative Neck Loading Experimentation**

Non-injurious neck loading is an important part of the neck biomechanics literature. This section summarizes applicable human subject research. However, experimental documentation and data in the literature are rather sparse for a variety of reasons. Approval difficulty, cost, and the need for specialty equipment involved in testing human subjects are among these reasons. The need for specialty equipment implies the need for a centralized resource, which has been centered in DoD facilities. Unfortunately, these organizations often do not have the resources or the approval authority to provide results into the open literature beyond a technical report summarizing the DoD specific results from individual studies. The majority of human subject testing with and without head supported mass has been performed by either the Air Force Research Laboratory (from the 1970s to the present) or the Naval Biodynamics Laboratory (NBDL) (from 1972 to the early 1990s when they ceased human subject testing) (NBDL, 1993).

Horizontal and vertical accelerative test sleds upon which these laboratory experiments are carried out necessitate that experiments are performed in a single axis of acceleration; primary accelerative inputs include +/- Gx (rear/frontal impact), Gy (side impact), or +Gz (vertical impact). Additionally, subjects can be oriented supine on the horizontal accelerator to observe response to –Gz acceleration. **Table 2** provides a general summary of human biodynamic response to accelerative input based upon video analysis of head kinematics. As shown, the human head often undergoes substantial rotational acceleration in response to linear acceleration. Further, human biodynamic response is highly variable in the dynamic accelerative testing environment; therefore it is important to conduct human subject testing to observe and record human neck response.

**Table 2. Human Biodynamic Neck Response to Accelerative Input**

| Accelerative input | Primary cervical spine biodynamic response |
|---|---|
| +Gx (rear) | Extension (-My) |
| -Gx (front) | Flexion (+My) to combined flexion (+My) and axial tension (+Fz) |
| Gy (side) | Combined twisting (Mz), side bending (Mx), and flexion (+My) |
| +Gz (vertical) | Axial compression (-Fz) with transition to slight flexion (+My) |
| -Gz (supine) | Axial tension (+Fz) |

In a foundational aviation-specific, ejection related study, Buhrman and Perry from the Air Force Research Laboratory (AFRL) conducted initial tests on the biomechanical effects of ejection acceleration on neck compression, shear, and bending moments under the load of helmets ranging from 1.45 kg to 3.0 kg (Buhrman and Perry, 1994). These tests were performed on AFRL's Vertical Deceleration Tower (VDT), the only man-rated human subject testing apparatus like it in the world. Subjects were seated and restrained as a pilot would be in an ejection seat representative of those used in current fighter aircraft. Using human subjects and a 97th percentile ATD, researchers collected data in a +Gz environment at variable helmet weights with constant acceleration as well as with constant helmet weights at variable, increasing,

acceleration. While these tests involving human participants were performed at accelerations below those often experienced during an ejection to maintain a minimal risk environment, results demonstrated that the mean of the peak compressive and shear neck load values recorded for each test, as well as mean peak neck bending moment, typically increased linearly with increases in acceleration at constant helmet mass. Similarly, these three measurements all generally increased linearly with increases in helmet mass at a constant +10 Gz acceleration. In the ATD tests, compression loading exceeded cadaver injury limits published by Mertz and Patrick during the test at +15 Gz with a 2 kg HMD (Mertz and Patrick, 1971). Thus they concluded that based upon known injury limits for compression, and a +10 $G_Z$ acceleration, total helmet mass should be kept under 2 kg to prevent injury to pilots (Buhrman and Perry, 1994). This study was one of the first to establish a rudimentary criterion for head supported mass of HMDs to avoid excessive neck loading during pilot ejection. This research was part of the initial work that established an interim criterion that provides rudimentary design guidelines for HMD weight and CG. Details of this interim criterion are provided in a later section of this chapter (entitled The Knox Box).

Follow-on studies considered neck response to lateral acceleration (Gy), testing HMDs of 1.36 and 2 kg and lateral accelerations of 4, 5, and 6 Gs, which resulted in a linearly increasing relationship between x and y direction shear neck loads, and head moments about the x and y rotational axis (relative to the OCs), and increased acceleration (Perry et al., 2003). Test subjects were seated and restrained in a representative ejection seat attached to a horizontal acceleration sled and oriented such that the sled accelerated the subject in a sideways manner down the test track. The study reported that the prominent y-axis angular acceleration of the head under these conditions was forward flexion.

Others have conducted research comparing male and female subjects in accelerative tests to expand the field of knowledge relevant to the smaller end of the anthropometric spectrum to ensure this population was not put at undo risk as a result of heavier HMDs (Perry, 1998;

Buhrman and Mosher, 1999; Buhrman et al., 2000; Buhrman and Wilson, 2003). Perry (1998) observed that female neck response overall was similar to that of the males in the study and reasoned that based upon other research showing females had a 25% greater risk for fatality in dynamic environments, modifications should be made to the existing injury criteria (Perry, 1998). All four of these studies found that ejection injury criteria would need to be revised to accommodate the risk of injury to smaller crewmembers using HMDs due to the fact that all previous injury criteria did not include adult individuals at the lowest end of the anthropometric spectrum.

Typically these human subject studies have been performed at accelerations no higher than 8 to 10 Gs and with HMDs of less than 3 kg, which have been safe for the volunteer subjects. While neck pain and soreness may have been recorded, these levels have been proven to be generally sub-injurious. However, human subject experiments alone are not adequate to construct an injury risk curve since injurious data points are also required to perform the necessary regression to construct the appropriate risk curves. In summary, these previous studies have contributed to understanding human neck response to accelerative environments at sub-injurious levels with head supported mass. They have been essential to understanding the biodynamics of humans in accelerative environments as well as understanding thresholds for tolerable neck forces experienced by humans, both with and without HMDs. Non-injurious testing has also been critical to develop the low-risk portion of injury risk curves. The next section covers the importance of testing in the ranges of acceleration that would cause injury to humans using PMHS, ATDs, and animals.

**Accelerative Neck Loading Experimentation at Injurious Levels**

Injurious neck loading is another segment of the literature applicable to neck injury criterion development. Experimental evidence cannot be ethically collected using human

subjects at levels that are likely to result in injury. Therefore studies at injurious levels are typically performed using a human analog. The most common are ATDs and PMHSs. Neck loads that cause injury, and the associated human neck limits to force exposure, have been determined in research associated with civilian biomechanics research, the automotive industry, and civilian and military aviation using these analogs. ATDs are used in automotive testing as well as aviation ejection testing where accelerative forces are known to be unsafe for human subject testing. The benefit of using ATDs is they are able to represent human movement and measure loads imposed by the dynamic accelerative environment. In laboratory, single axis accelerative sled tests at AFRL (previously mentioned), where human subjects are being tested, standard protocol is to first perform the test on an ATD to observe the forces to ensure it is safe for humans. In automotive standard tests to evaluate restraint systems, ATDs are used because it is known that the impact can be injurious to humans.

The basis for the current neck injury criterion in the United States for determining safety of new automotive crash restraint equipment (Nij, discussed in the next section), comes from research performed at injurious levels by Prasad and Daniel and Mertz et al. (Prasad and Daniel, 1984; Mertz et al., 1997). These researchers performed a series of matched frontal crash tests comparing piglet injuries and associated ATD neck loads. The piglets were used to determine a neck injury criterion for a 3-year-old child based upon their similarity in size, weight, and state of tissue development. The representative ATD provided the neck forces for the crash test and an autopsy of the piglet determined the level of injury. A neck injury criterion for a 3-year-old child was estimated based upon the results, and a neck injury criterion for other size occupants was constructed based upon scaling factors (Eppinger et al., 1999; Mertz et al., 1997). Neck injury risk curves were then constructed by analyzing non-injurious and injurious loading using logistic regression (Eppinger et al., 1999). Follow on PMHS studies to continue to determine the

strength of the cervical spine in flexion and extension to improve ATD design and risk function models have been performed (Nightingale et al., 2002; Nightingale et al., 2007).

Similar to automotive testing, ejection seat testing employs ATDs to qualify new ejection seat programs and additions or modifications to the ejection seat system (i.e., HMD addition or seat modification) to ensure pilot safety standards are met. These tests are performed over the spectrum of potential accelerative forces, and thus an ATD is preferred when evaluating the forces involved. For DoD aviation systems, these evaluations are performed on rocket sled tracks at Holloman Air Force Base, New Mexico and at Langford Lodge, Ireland. Instrumented ATDs are placed in the test seat and restrained as a pilot would be restrained. Then they are accelerated down the sled track and the seat deploys as it would in a real-world, in-air ejection. Neck loads as well as loads to other parts of the body are recorded throughout the duration of the ejection sequence. Of specific interest to an improved neck injury criterion that meets the AFLCMC requirement to be multi-axial are the six major neck loads; neck shear (*Fx, Fy*), neck tension/compression (*Fz*), anterior/posterior bending moment (*My*), coronal moment (side bending – *Mx*), and twisting (*Mz*). Qualification protocol calls for static testing of the escape system as well as 18 sled tests. Four of the tests are conducted each at minimum and maximum air speed with the additional 10 tests conducted at intervals between minimum and maximum speeds to ensure maximum data on escape system performance for new escape system development (MIL-STD-846C, 1974). Additionally, for reliability, the escape system is required to successfully complete an additional four tests (for a total of 22). Representative plots of the six major neck loads recorded from an instrumented 145 lb ATD during an ejection sled test at 227 KEAS are provided in **Figure 5** and **Figure 6**. The major ejection phases are labeled in the time sequence.

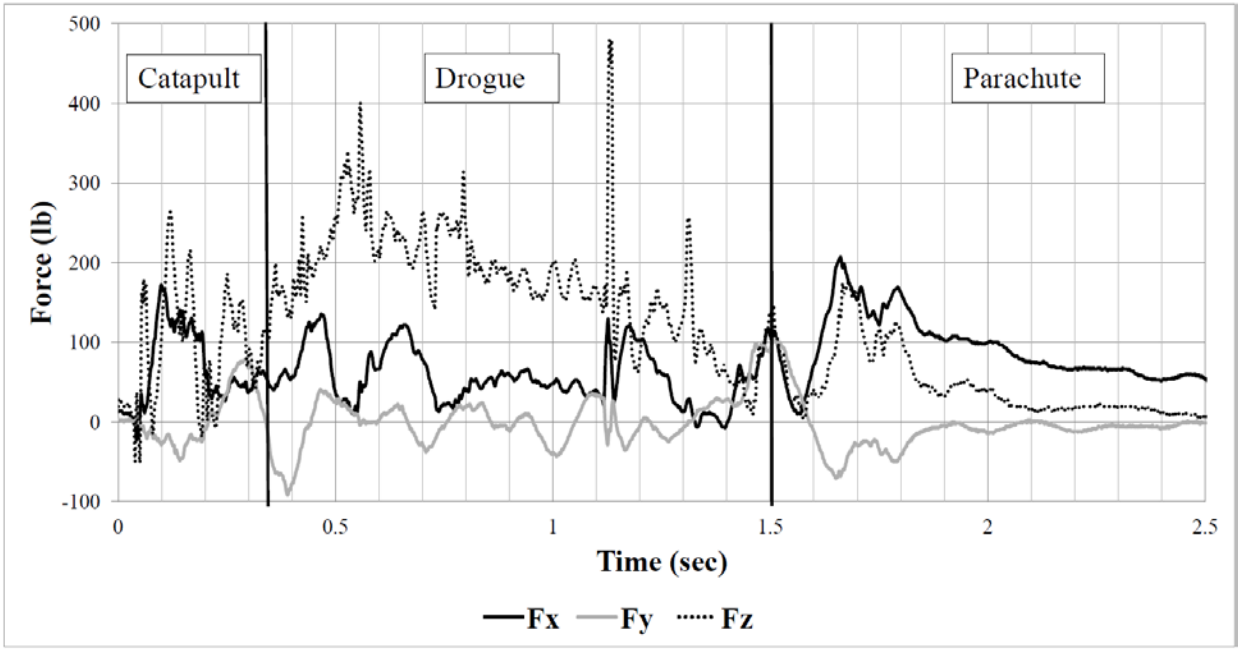


**Figure 5. Time History of ATD Upper Neck Forces During Ejection Seat Sled Test**

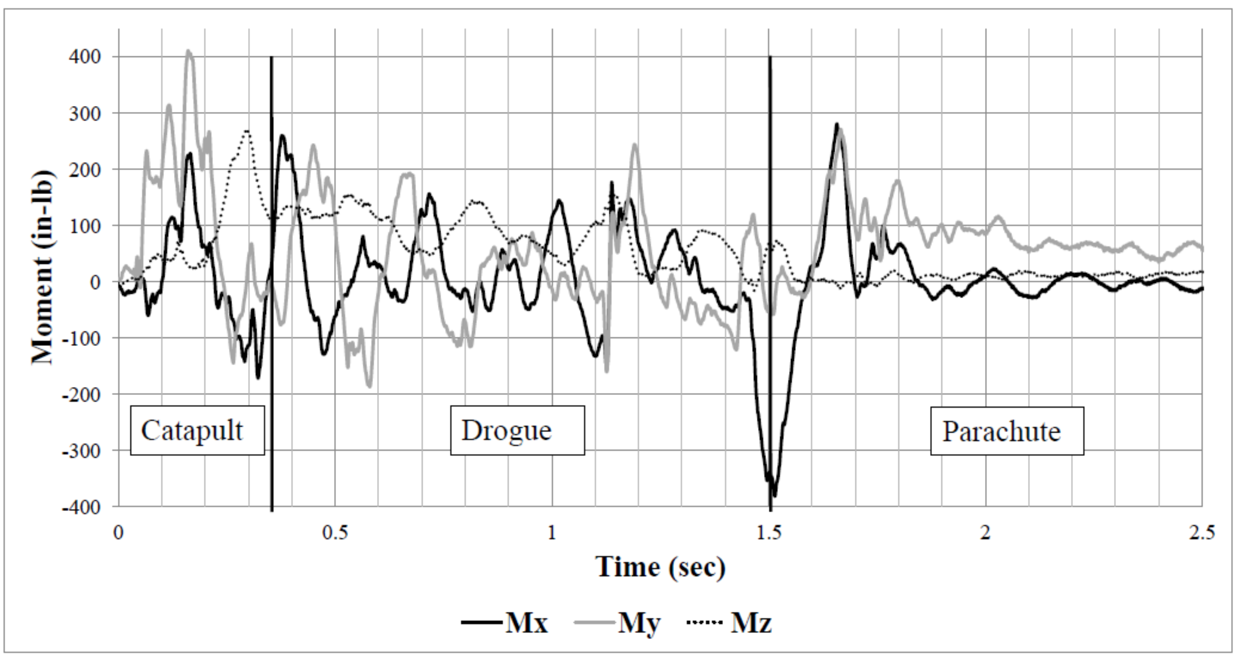


**Figure 6. Time History of ATD Upper Neck Moments During Ejection Seat Sled Test**

Many civilian biomechanics research programs have conducted studies with PMHSs to further knowledge of the limits of human tissue, injury causality mechanisms, and kinematics of

PMHSs in accelerative environments. This literature has served to further the body of knowledge in injury prevention in automotive and military environments. Ivancic and Sha evaluated neck injury criteria during simulated rear-end collisions using six fresh, frozen whole cervical spine specimens (Ivancic and Sha, 2010). The specimens were used to construct a physical model of a human upper half; an anthropometric surrogate head was attached to the spine, and the spine was imbedded into a rear-impact dummy torso. Others have also used PMHS to better understand and investigate the limits of the human cervical spine in ways that cannot be performed with human subjects (Panjabi et al., 1998; Eichberger et al., 2000; Stemper et al., 2003; Kettler et al., 2006; Ivancic and Xiao, 2011; Yoganandan et al., 1996). The details of these studies are beyond the scope of this AF neck injury criteria focused research, but to the extent that they provide applicable data they will be consulted. To summarize, this body of research has aided the determination of limits of the strength of the human cervical spine.

## Neck Injury Criteria

This section provides an overview of the pertinent neck injury criteria that exist today in various fields of application. These criteria have been developed to protect humans from injury in various accelerative environments. Understanding existing neck injury criteria is an important step in the process of developing aviation-specific, ejection neck injury criteria, as portions of these criteria may be helpful and valuable to include in updated criteria developed in this work.

In general two types of neck injury criteria exist. The first type limits peak instantaneous neck loading and is the most common among the criteria reviewed in this work. The second type of criteria limits the time duration of neck loading at specific input levels. A graphical example of a peak instantaneous neck injury criterion is shown in **Figure 7**. It shows a time history of the $N_{ij}$ observed in the neck load cell of an ATD during testing of an escape system on the rocket sled track at Holloman Air Force Base. The limit of peak instantaneous neck load is an $N_{ij}$ of

0.5, and it can be seen that at a time close to 0.3 seconds an exceedence was observed. However, this exceedence lasted for only a few milliseconds.

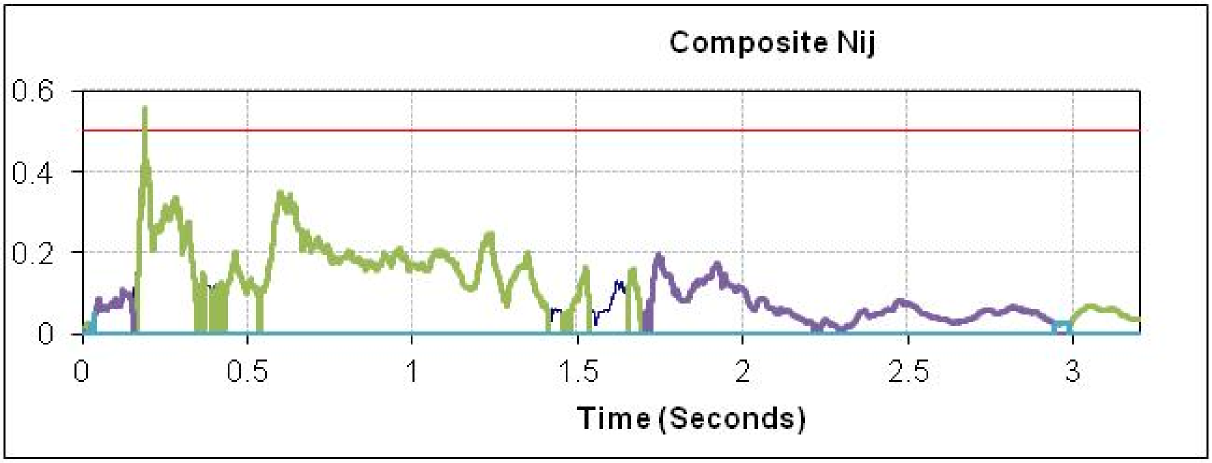


**Figure 7. Example Sled Test Upper Neck ATD Data Compared to Nij Limit of 0.5**

**Figure 8** depicts a graphical example of a typical load duration limit. This specific example is the load duration limits for upper neck tension, compression, and shear used in the NIC adapted for the ejection environment from the Mertz duration limits developed for the automobile industry, which will be discussed later. For each specific loading type (tension, compression, or shear), the upper dashed line depicts the load magnitude and duration limit. This example uses an ATD to measure the neck loads during an ejection system test sequence at the Holloman AFB rocket sled test track. The lower solid line represents the measured neck load magnitude and duration observed in the ATD neck. The specific test illustrated in **Figure 7** passed the duration limit for each load type, as the observed magnitude and duration of the ATD neck load was below the limit set forth by the criterion.

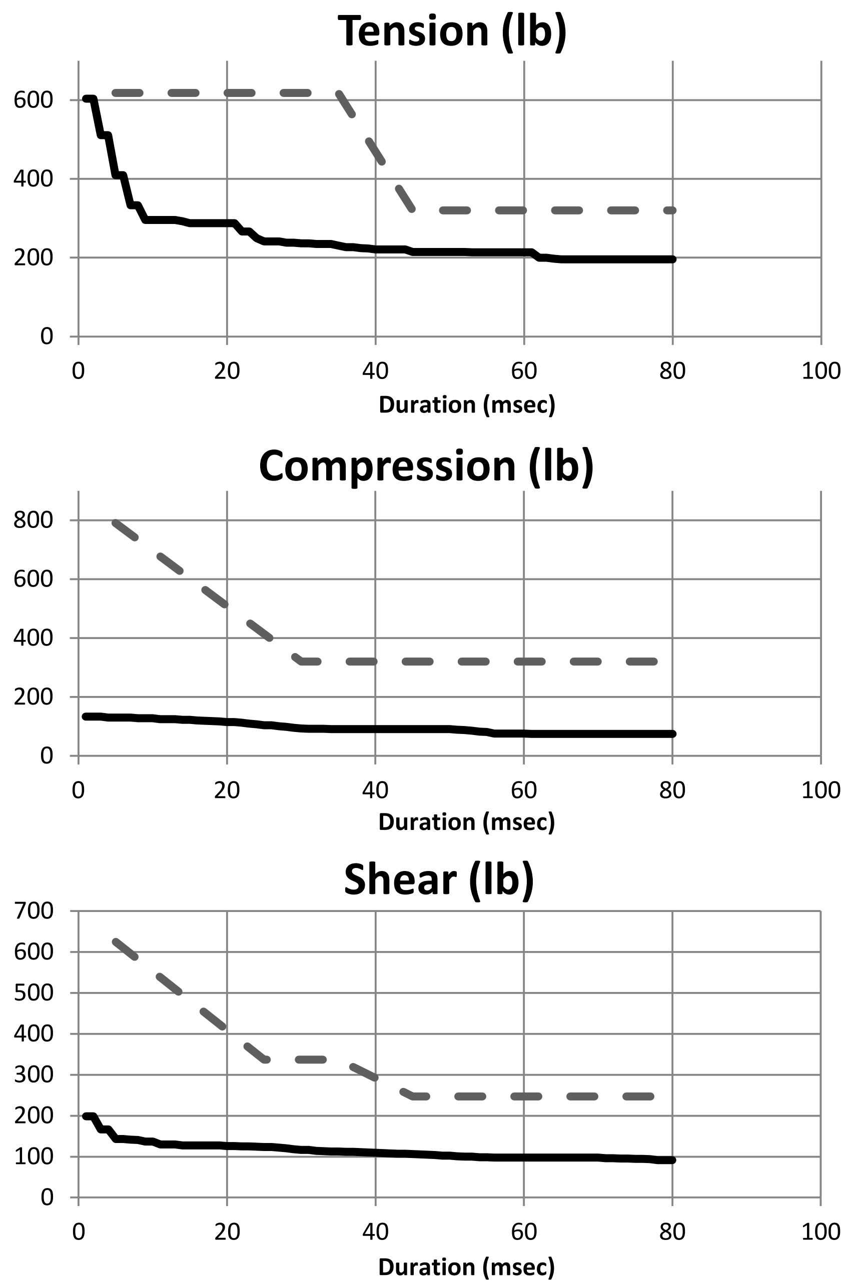


**Figure 8. Example of Upper Neck Load Duration Criterion Used in NIC**

### Injury Classification

Injury classification is an important topic to highlight when discussing injury criteria. In general, injury criteria are developed with a defined level of injury. In general, a minor injury

involves only the soft tissue, with no bone fractures (Bogduk and Yoganandan, 2001). A major injury involves a fracture of the cervical spine or a neurologic injury that involves either the spinal cord or the nerve roots (Cusick and Yoganandan, 2002). More specifically, the Abbreviated Injury Scale (AIS) is a clinical index of injuries that specifically defines the injury and assigns it a severity rating from 0 to 6 (AAM, 2008). The AIS is commonly used in many injury criteria currently employed due to its exact delineation of the type of injury and its corresponding severity. These features make the AIS ideal for use when generating injury risk curves at specific AIS levels for the purpose of limiting the injury. While the AIS specifically classifies injury in detail and labels each with a severity, in general AIS 1 is minor, AIS 2 is moderate, AIS 3 is serious, AIS 4 is severe, AIS 5 is critical, and AIS 6 is maximal. The following subsections provide detail of pertinent neck injury criteria.

### Mertz Criteria

Mertz developed neck tension, compression, and shear force duration criteria for the automotive industry which has evolved over the years. This research was initially based upon studies comparing Hybrid III (a specific ATD used in automotive and other environments) neck response to simulated tackles that resulted in injuries to football players. These criteria were then updated for application to the automobile industry to include scaling for occupant size, muscle activation, and multiple loading directions (Mertz, 1993; Mertz et al., 1997; Armenia-Cope et al., 1993). The resulting injury assessment curves for these criteria are displayed through graphs delineating maximum allowable cervical spine tension, compression, and shear loads for the upper (OC) and lower (C7-T1 junction) neck for each time duration. Details of the loads and duration limits in graph and tabular form are available in the literature (Nichols, 2006; Paskoff and Sieveka, 2004; Carter et al., 2000). An example graph of the Mertz neck tension

force duration limits based upon occupant size is provided in **Figure 9**. Similar graphs exist for the compression and shear force portion of the injury criteria, and tables are available for each mode of neck loading as well.

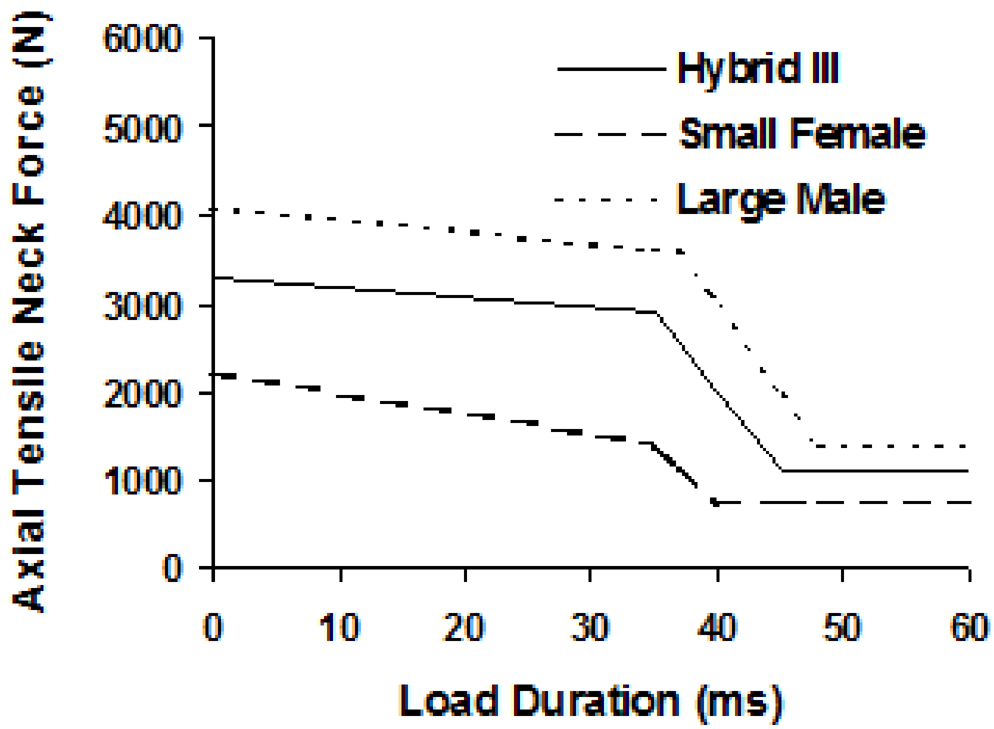


**Figure 9. Mertz Tensile Neck Force Duration Criteria** (Mertz, 1993)

The basis of this sustained force neck injury criterion is the hypothesis and observation that the neck can sustain higher loads for shorter periods of time and lower loads for an extended duration of time. The magnitudes of these load durations determine the threshold above which there exists a potential for significant neck injury and below which significant neck injury is unlikely. These thresholds were established through human and PMHS experiments. Also, a variation of these duration limits were adapted by the US Navy for use in the aviation domain and are a part of an overall aviation neck injury criteria called the NIC, which will be discussed later (Nichols, 2006).

### $N_{ij}$ Criteria

The following paper, entitled "Evaluation of the $N_{ij}$ Neck Injury Criteria with Human Response Data for Use in Future Research on Helmet Mounted Display Mass Properties," was published in the Proceedings of the Human Factors and Ergonomics Society 56th Annual Meeting (Parr et al., 2012).  It provides a background of a widely used neck injury criterion called the Nij, as well as a preliminary assessment of the potential application of the criterion to the aviation environment.  The paper is included in its entirety starting on the next page.

### *Abstract*

Technological advances have enabled components to be added to Helmet Mounted Displays (HMDs) that provide increased pilot capability. Future Air Force fighter aircraft are being developed to incorporate added technologies that could result in heavier and bulkier HMDs. The added weight and center of gravity changes to the pilot's helmet ensemble from these additional components place the neck at an increased risk of injury during ejection. This paper outlines a preliminary research methodology studying the human neck response data from the Air Force Research Laboratory's extensive human impact testing database using the $N_{ij}$ criteria as an evaluative tool. Initial results are presented.

### *Introduction*

Helmet Mounted Displays (HMDs) are becoming common human-machine interface equipment in manned flight. They have been designed to increase the performance of operators in their weapon system and thus increase overall mission effectiveness (Rash et al., 2009). Currently in use on multiple Department of Defense (DoD) weapon systems (e.g., F-15, F-16, F-18 and planned for the F-35 Joint Strike Fighter), HMDs add capability, enable faster data processing and information fusion, and enhance mission accomplishment across the spectrum of military operations. However, important parameters must be carefully implemented into HMD design to ensure pilot safety is preserved. The addition of added weight through the inclusion of night vision goggles, targeting displays, and other components to the existing pilot helmet ensemble has the potential to increase the risk of operator neck injury if the pilot is subjected to accelerative environments such as ejection from the aircraft. Injury due to a heavier HMD with a highly off-center center of gravity (CG) in this environment could range from low severity strains and muscle tears, to high severity cervical spine fractures and ligament ruptures. Design parameters affecting pilot neck biomechanics that should be considered include minimizing the

weight of the HMD and distributing the mass of the components of the HMD such that the center of gravity remains as close to that of the head alone (Melzer, 2001). However, there is a tradeoff between adding additional functionality to improve the pilot's likelihood to survive in combat and additional weight which could lead to operator fatigue and potential injury during ejection. It is therefore important that pilot neck response due to heavier HMDs be understood and characterized using a standard evaluation criteria that could be applied during early concept definition and design.

The National Highway Transportation Safety Administration (NHTSA) has established a neck injury criteria called the $N_{ij}$ that the auto industry must follow within the United States as part of a comprehensive crash protection safety standard used in the assessment of advanced automotive restraint systems (Eppinger et al., 1999; Eppinger et al., 2000). The primary purpose of this criterion is to provide a consistent and quantitative method for evaluating and differentiating automotive crash and restraint systems where the quantitative metric (e.g., $N_{ij}$) can be related to the likelihood of injuries in specified severity categories. This metric has a strong foundation in biomechanics and relies upon results of crash tests with standardized Anthropomorphic Test Devices (ATDs) to provide a criteria for predicting the likelihood of injury to persons with varying anthropometric characteristics for various automotive crash and restraint systems (Eppinger et al., 2000). The ability to define a relationship between the performance of the automotive crash and restraint system and the likelihood of injury, especially for persons with varying anthropometric characteristics, is a key attribute of the $N_{ij}$ criteria that is highly desirable and that does not exist for any known helmet-mounted display evaluation system.

To analyze the characteristics of human neck response using the $N_{ij}$ model, data from an extensive repository collected by the Air Force Research Laboratory's (AFRL) Warfighter Division (711$^{th}$ HPW/RHC) was used. This organization maintains a human test database of

neck response under various accelerative and head loading conditions collected from studies using AFRL's Vertical Deceleration Tower (VDT) and Horizontal Impulse Accelerator (HIA) facilities. These are the only known, human-rated impact testing facilities in the world. The purpose of this paper is to outline a preliminary research methodology for studying the human neck response data from the AFRL database using the $N_{ij}$ criteria as an evaluative tool. Using this standard NHTSA $N_{ij}$ model to evaluate existing neck response data will hopefully provide insight to the following research questions:

1) What $N_{ij}$ values emerge from the non-injurious, variable HMD weight and CG, accelerative environment human data?
2) Can an aviation, ejection-specific, human data supported neck injury criteria, similar to the $N_{ij}$ be developed to facilitate design of the mass properties of future HMDs and evaluate risk posed by various HMD configurations?

### *Background*

The foundational research of neck injury thresholds, tolerance to impact, strengths and biomechanical properties of biological materials, and injury pathways was initially accomplished for use by the automotive research community (Mertz and Patrick, 1971; Yamada, 1973; Sances et al., 1981; Brinn et al., 1986). The military aviation community began designing and building ejection seats for high speed aircraft after World War II. As pilot safety became increasingly important and additional head supported mass (helmets, oxygen masks, etc.) became common during the Vietnam War and following, neck injury risk took on greater importance to the Air Force operational and research communities. Aviation-specific research expanded the understanding of human response to highly accelerative, ejection-like environments, performed at the Air Force Research Laboratory (AFRL) on ATDs, human subjects and post mortem human

subjects (PMHS) to understand the full range of biomechanical effects, neck response, and injury pathways and thresholds on human neck muscles, ligaments, tendons, and vertebrae.

When a pilot ejects from an aircraft there are four different phases which subject the pilot's head and neck to different accelerative forces. In order, these phases are catapult stroke, windblast, seat stabilization, and parachute opening shock. Most aviation-specific ejection studies have been done on the effects of the catapult stroke, in which the accelerative forces act upon the head and neck in the positive z axis (upward, or $+G_z$, see **Figure 4** for anatomical coordinate system). As added helmet weight was introduced into AF operations in the form of helmet mounted equipment like HMDs and night vision goggles (NVGs), numerous studies evaluated ATD and human neck response to various head supported mass and various CGs (Buhrman and Perry, 1994; Perry, 1994; Perry and Buhrman, 1995; Perry and Buhrman, 1996; Perry et al., 1997; Perry, 1998; Buhrman and Wilson, 2003; Salzar et al., 2009). Other research has evaluated neck response from other phases of ejection, which include exposure to frontal ($+G_x$) and sideward ($+G_y$) acceleration (Buhrman et al., 2000; Perry et al., 2003; Doczy et al., 2004).

In a foundational aviation-specific, ejection related study, Buhrman and Perry conducted initial tests on the biomechanical effects of ejection acceleration on neck compression, shear and bending moment under the load of helmets ranging from 1.47 kg to 3 kg. Using human subjects and a 97$^{th}$ percentile ATD, researchers collected data in test configurations with variable helmet weights and constant acceleration as well as test configurations with constant helmet weights and variable acceleration (Buhrman and Perry, 1994). Tests were performed using the VDT at Wright Patterson AFB. Results from this extensive study showed that, in general, compressive and shear neck load, as well as neck bending moment linearly increased with increases in acceleration forces at constant helmet weights. Similarly these three measurements all generally increased linearly with increases in helmet weights at a constant +10 $G_z$ impact acceleration.

They concluded that based upon known injury limits at the time, total helmet weight should be kept under 2 kg to prevent injury and neck fatigue to pilots (Buhrman and Perry, 1994). This study was one of the first to call for criteria to be established for neck loading under the higher head supported mass of HMDs.

Follow-on studies considered neck response to lateral impact acceleration, testing HMDs of 1.36 and 2.04 kg and lateral accelerations of four, five, and six Gs, which resulted in a linearly increasing relationship between neck loads and moments and increased acceleration (Perry et al., 2003). Others have accomplished research comparing male and female subjects in impact tests to expand the field of knowledge relevant to the smaller end of the anthropometric spectrum to ensure this population was not put at undo risk as a result of heavier HMDs (Perry, 1998; Buhrman and Mosher, 1999; Buhrman et al., 2000; Buhrman and Wilson, 2003). All four of these studies found that ejection injury models would need to be revised to accommodate the risk of injury to smaller crewmembers using HMDs. In sum, this research has contributed to understanding human neck response to accelerative environments to protect pilots. However, this past research has focused on single neck force values; combined loading results were not accounted for and an aviation-specific, robust, neck injury criteria has yet to emerge to ensure pilot neck protection with heavier HMDs in accelerative environments.

### *Neck Injury Prevention Criteria*

NHTSA's neck injury criteria, the $N_{ij}$, established critical limits in four types of neck loading that NHTSA engineers determined to be dominant in automotive crashes; axial loading (tension and compression), and sagittal plane bending moments (flexion – forward, and extension – rearward) using a methodology initially presented by Klinich et al. (Klinich et al., 1996). The researchers who developed these injury criteria applied previous biomechanical research experiments using volunteer humans, porcine subjects and PMHSs and determined the

combinations of these four types of neck loading to be most important when evaluating neck injury in frontal crashes. This same research established critical limits for these four load pathways (Mertz et al., 1978; Nyquist et al., 1980; Mertz and Patrick, 1971; Yoganandan et al., 1996; Shea et al., 1992; Lenox et al., 1982).

The formula used to calculate the $N_{ij}$ is;

$$N_{ij} = \frac{F_Z}{F_{\text{int}}} + \frac{M_Y}{M_{\text{int}}} \tag{1}$$

In this equation, $F_Z$ and $M_Y$ are specific to an automotive crash and restraint system under evaluation. $F_Z$ is the maximum measured axial load in tension or compression and $M_Y$ is the maximum measured flexion or extension bending moment. Each of these values are determined from calculations involving the mass of the automotive occupant's head and the maximum acceleration of a head, typically the head of a crash dummy, during a standardized automotive crash scenario. The values $F_{int}$ and $M_{int}$ are critical load values established by the NHTSA for the maximum axial load in tension or compression and the measured flexion or extension bending moment established by NHTSA (Eppinger et al., 1999). Different critical load values are established for groups of individuals within different anthropometric categories as shown in **Table 3**. The "ij" subscript of the $N_{ij}$ signifies indices for the four combination mechanisms for injury, $N_{TE}$, $N_{TF}$, $N_{CE}$, and $N_{CF}$, where T and C represent the axial load index (tension or compression) and F and E represent the sagittal plane bending moment index (flexion or extension) (Eppinger et al., 1999). The current $N_{ij}$ "performance limit" is set at 1.0, meaning an automotive test that produces ATD neck loads that exceed an $N_{ij}$ value of 1.0 fails the criteria. An $N_{ij}$ of 1.0 represents a 22% risk of a greater than 3 Abbreviated Injury Scale (AIS) injury, considered a moderate injury (Eppinger et al., 1999). The risk curves associated with $N_{ij}$ values

are an important part of the criteria as they provide likelihood of injury information and are covered in detail in Eppinger et al. (Eppinger et al., 1999).

**Table 3. Critical Intercepts for the Nij (from Eppinger et al., 2000)**

| | Large Sized Male§ | Mid-Sized Male | Small Sized Female |
|---|---|---|---|
| **Neck Criteria**: Nij | 1.0 | 1.0 | 1.0 |
| In- Position Critical Intercept Values | | | |
| Tension (N) | 8216 | 6806 | 4287 |
| Compression (N) | 7440 | 6160 | 3880 |
| Flexion (Nm) | 415 | 310 | 155 |
| Extension (Nm) | 179 | 135 | 67 |
| Peak Tension (N) | 5030 | 4170 | 2620 |
| Peak Compression (N) | 4830 | 4000 | 2520 |

§ The Large Male ($95^{th}$ percentile Hybrid III) is not included in the final rule, but the performance limits are listed here for informational purposes.

The use of the $N_{ij}$ in the aviation community is not completely new. Researchers in past ejection seat testing at AFRL have recorded and computed $N_{ij}$ values in ATD testing to ensure conditions for subsequent human testing under the same conditions were sub-injurious. However, to the authors' knowledge it has not been evaluated, qualified, or verified using human neck response data for the purposes of using the results as an evaluative tool for HMD design. Based upon recommendations from the AFRL Warfighter Division ($711^{th}$ HPW/RHC) and the Naval Air Systems Command (NASC), the aviation community has proposed an $N_{ij}$ performance limit of 0.5 rather than NHTSA's 1.0 limit (Nichols, 2006). This lower limit provides a reduced risk of injury, which is essential because of the unique requirements of military aviation. The

lower limit was proposed because a pilot could potentially be required to evade capture or navigate to an extraction point to be rescued by a combat search and rescue team to successfully survive an ejection event. NHTSA’s performance limit of 1.0 is acceptable in the automotive environment because of the assumption that first responders will be on site shortly after a car accident to attend to any injury sustained in the collision, an assumption that cannot be made in an ejection scenario.

The two component factors of the $N_{ij}$ (axial load and bending moment) can also be plotted, which provides a visual representation of acceptable neck loads. For example, the critical intercepts for the neck injury criteria as shown in **Table 3** for a mid-sized male can be plotted to form the kite-shaped region in **Figure 10**. $N_{ij}$ values recorded from an automotive test crash and restraint system can then be plotted within this figure, showing the combined forces and moments plotted each millisecond of the duration of the impact. Acceptable force values would lie inside the region of **Figure 10**, while unacceptable values would fall outside of the shaded region.

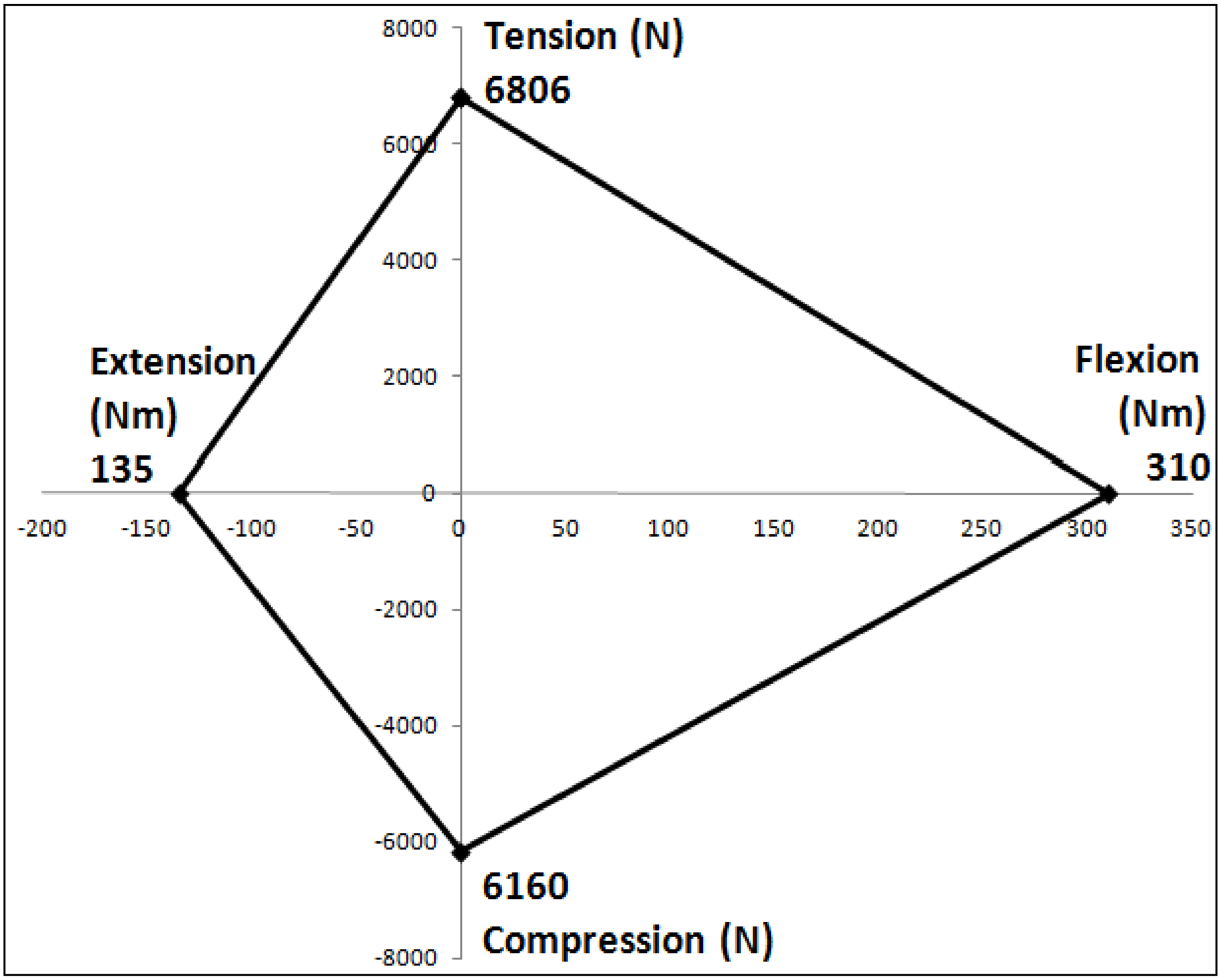


**Figure 10. Nij = 1.0 Neck Injury Criteria for 50th Percentile Male Dummy (Eppinger et al., 2000)**

The $N_{ij}$ was formulated specifically for ATD testing. To test a new automobile restraint system the ATDs listed in **Table 3** are placed in the vehicle with the restraint system in place. They are fit with instrumented necks which have been designed to simulate the representative human neck (i.e. child, female adult, male adult, etc.). Forces collected from these instrumented necks during the crash tests are used to calculate the $N_{ij}$ and determine if the test, and thus the restraint system, passed or failed. It is important to note that $N_{ij}$ does not give restraint system designers information to guide design. Instead, it provides a pass/fail evaluation of a fully prototyped restraint system.

### *Methods*

In this study, an initial set of data from an experiment on the effects of variable helmet weight on human response to -$G_X$ impact was plotted using the $N_{ij}$ criteria to characterize non-injurious neck response to impact. The preliminary set of data used in this analysis was from an experiment in which 8 Gs of accelerative input was applied to 23 human subjects wearing an HMD weighing 2 kg. This horizontal testing was intended to simulate the forces experienced by crewmembers in the seat deceleration and parachute opening shock phases of ejection. This mode of testing also mimics most closely a frontal collision test used to gather $N_{ij}$ data when testing a new restraint system.

During the test, volunteer subjects were seated vertically and restrained in a standard Air Force fighter aircraft ACES-II ejection seat. The seat was mounted to AFRL's HIA and subjects were accelerated backwards at the appropriate acceleration level to measure the -$G_X$ neck responses. The accelerative portion of the experiment lasted for about 200 ms and data were collected every millisecond. All of the tests were non-injurious but neck pain was reported 20% of the time, mostly at the higher helmet weights and acceleration levels. For further details of the experimental set up, methods, and results the reader is referred to Doczy et al. (Doczy et al., 2004).

Neck load data were plotted using the $N_{ij}$ model and analyzed. Independent variables for this research included helmet weight and acceleration applied to the test sled. The dependent variables were resultant head, neck and body accelerations which were used to compute neck loads (shear, tension, compression, flexion, and extension). It should be noted that the program used to calculate the neck loads included a small addition to the bending moment values based upon an offset for the occipital condyles inherent in the program for calculating ATD neck loads. Since this addition is not necessary when calculating human neck loads it is possible that the

bending moments are slightly overstated. This will be corrected in future data analysis and neck load calculations. It does not affect tension or compression values.

This initial data set analyzed only the neck response from horizontal $-G_X$ acceleration. In follow-on work, additional data sets will be evaluated from the results of extensive experiments testing vertical ($+G_Z$) and lateral ($-G_Y$) impacts with helmets up to 3 kg with variable CGs, and acceleration inputs of up to 10 Gs.

### *Preliminary Results and Discussion*

For each of the 23 test runs on human subjects a time history of the observed $N_{ij}$ was plotted. The $N_{ij}$ time history is the recorded $N_{ij}$ value at every millisecond of the 200 ms data set. The $N_{ij}$ value at any one time will fall only into one quadrant, depending on the combination of neck forces experienced by the subject in that instant ($N_{TE}$, $N_{TF}$, $N_{CE}$, and $N_{CF}$). The $N_{ij}$ was calculated and plotted to show where these values fell on the plot in relation to the NHTSA mid-sized male intercept values for an $N_{ij} = 0.5$, the limit proposed for use in the aviation environment. The initial data plotted in the $N_{ij}$ format are shown in **Figure 11**.

The "point cloud" generated by the plot is significant because it is a space within the $N_{ij}$ where humans have been safely tested. The $N_{ij}$ is normally used to show the fail zones of a test, but in this case the human data provides a known safe zone. This provides important information at the non-injurious level of the spectrum, which can be used for future HMD design. To the authors' knowledge this is the first time the $N_{ij}$ has been plotted with actual human response data.

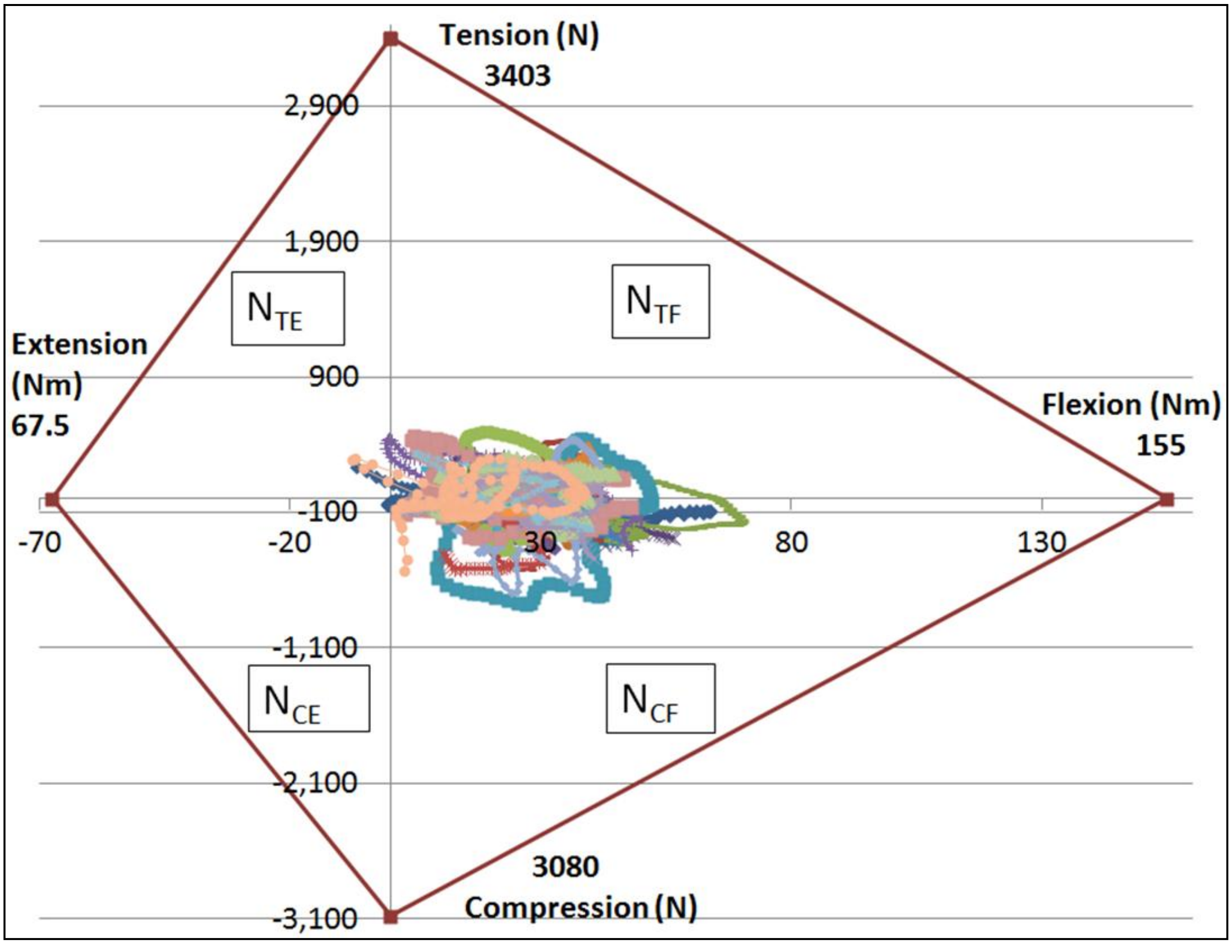


**Figure 11. Plot of Human Nij Values and Intercepts (8 G Acceleration with 2 kg HMD)**

As expected, all of the neck loads from this human subject experiment were well within the aviation field's proposed $N_{ij}$ performance limit of 0.5. The highest $N_{ij}$ value observed was 0.22. The extension values were artificially lower due to the restraining effects of the test ejection seat headrest. This accounts for the limited data points in the $N_{TE}$ and $N_{CE}$ quadrants. The $N_{ij}$ values primarily fall in a central area of the cloud, but there are numerous values that approach the boundaries of this region. The variability of these $N_{ij}$ values are likely to come from a variety of sources; gender, neck strength, head size, neck length, and body weight. Understanding this variability requires further exploration; with more data points a robust statistical analysis can be accomplished. Future work will explore the nature and potential causes of the observed variability in greater detail. Follow-on efforts are planned to compare $N_{ij}$

neck responses from different acceleration levels and orientations, helmet weights, and helmet CGs in order to expand the known safe zone.

The significant contribution of this paper is the evaluation of the $N_{ij}$ criteria in the aviation ejection field. The $N_{ij}$ criteria has many positive attributes and has the potential to be applied to the field of aviation ejection neck safety in a way that helps develop and field HMDs with safe mass properties. It is likely that the robust biomechanical underpinnings of the criteria that have given it wide acceptance in the automotive community could translate cross-domain and be used as a tool to evaluate the impact of different HMD loading conditions and different accelerations applied in testing new HMDs on pilot neck safety. The criteria takes into account combined neck loading as well as providing injury risk curves associated with specific observed $N_{ij}$ values. Some limitations of the $N_{ij}$ were found as well. The downside of the $N_{ij}$ criteria for use in guiding HMD mass property design is that, as in the car industry, it provides reactive rather than proactive assessment of a system. The criteria will provide a safe or unsafe neck response assessment for a fully prototyped HMD, but in its current format does not provide design parameters to inform the early phase design process. Future research will focus on combining the $N_{ij}$ criteria with modeling techniques in a way that could provide estimated resultant $N_{ij}$ values based upon specific HMD mass properties. This would allow designers to design within acceptable neck safety parameters without having to produce a fully prototyped HMD to determine if it is a safe design or not.

### ***Conclusions***

This paper outlined a research methodology aimed at studying the human neck response data from the AFRL database using the $N_{ij}$ criteria as an evaluative tool. Preliminary data were presented and analyzed. It is expected that analyzing additional human data using the $N_{ij}$ model will lend further insight into the usefulness of the $N_{ij}$ as a neck injury criteria in the aviation

environment and further characterize patterns of human neck response in impact environments. The ultimate goal is to develop an ejection and aviation specific, human data supported model in order to more fully understand human neck response in accelerative environments to provide better safety criteria for the design of future HMDs.

### Lower Neck BEAM Criterion

Bass and colleagues proposed a neck injury criterion called the Beam Criterion for the lower neck based upon accelerative testing of PMHSs with head supported mass in various frontal and vertical orientations (Bass et al., 2006). Their lower neck injury criterion is structured similarly to the $N_{ij}$, based on a beam model of the lower cervical spine, though initially a shear component was included but later removed because it did not improve the predictive ability of the risk function. They tested 36 cadaveric head/neck complexes and six whole PMHSs under accelerative scenarios with varying head supported mass and acceleration. Additionally, the researchers performed accelerative experiments with Hybrid III and THOR (an advanced 50th percentile ATD with additional sensors) ATDs and six PMHSs. These experiments provided the injury and non injury data points, which produced a risk function and corresponding injury criteria. Injury level was investigated post test and AIS levels were determined for each specimen. The resultant risk function was constructed at the AIS 2 or greater injury level. It was observed in their specimens that injury to the lower neck was more prominent with the addition of head supported mass, and thus they constructed their injury criterion based upon forces at the lower neck (Bass et al., 2006). Additionally, rather than a logistic regression, they used a survival analysis method of regression to develop the risk curves in their study based upon the fact that their data set consisted of censored data; injury tests were left censored and non-injury tests were right censored (Bass et al., 2006). The critical values used as a starting point in the beam criterion to scale the axial loads and sagittal plane bending moments were taken from the NHTSA Nij 50th percentile male Hybrid III ATD simple bending values (4170 N tension, 4000 N compression, and 190 N-m flexion) (Bass et al., 2006). Once a baseline risk function was produced, the researchers determined optimum critical values by allowing the ratio between the flexion and tension critical values to vary and by constraining the mean 50% injury risk to equal 1.0 with standard deviation minimized (Bass et al., 2006). This

resulted in the risk function statistically optimizing the critical values (new values of 5660 N tension, 5430 N compression, and 141 N-m flexion) (Bass et al., 2006).

Bass et al. compared the Nij evaluated at the upper neck with their criterion evaluated at the lower neck and concluded that, based upon their experimental observations, the Nij was not an adequate neck injury criterion in inertial loading with head supported mass (Bass et al., 2006). This finding was based upon the fact that the overall kinematics of the Hybrid-III ATD was significantly different from cadavers in the accelerative testing with head supported mass. The authors posit that since the Nij is built around the neck response recorded from the Hybrid-III, the resulting neck injury conclusions drawn from a Hybrid-III test with head supported mass are flawed (Bass et al., 2006). It was observed that the THOR ATD had kinematics that more similarly matched the cadavers in testing. However, the bulk of the data involved in constructing the Beam Criterion were from PMHS segments potted such that they were only mobile from T2 and up (T3-T4 spinal segment was immobilized and potted into a mounting fixture). This may have caused the kinetic response to be different from a whole PMHS or ATD, potentially affecting the results. Salzar et al. found the Beam Criterion not to be accurately predictive of injury in small PMHS accelerative +Gz sled tests with head supported mass (Salzar et al., 2009) compared to the Nij and the NIC.

### USN Ejection Neck Injury Criteria (NIC)

A research team from the United States Naval Air Systems Command has put forth a set of neck injury criteria that is a set of metrics used to assess potential neck injuries in ejection, which will be referred to as the NIC. The researchers outlined the NIC used by the US Navy to assess the potential for pilot neck injury in ejection (Nichols, 2006). The purpose of these criteria is to field aviation systems that prevent neck injury hazards to pilots from escape systems. The criteria are used to qualify new equipment introduced into the ejection

environment and were most recently used to evaluate the Joint Strike Fighter (JSF) escape system in developmental testing from 2007 to 2010 and further testing starting again in 2014. These criteria have been employed to evaluate new ejection seat acquisition programs (e.g., JSF), ejection seat modification programs (e.g., the T/AV-8B Ejection Seat Improvement Program and Naval Aircrew Common Ejection Seat Stability Improvement Program), and HMD programs (e.g. F-18A/B Joint Helmet Mounted Cueing System) (Nichols, 2006). It incorporates 12 neck injury criteria, which include six modes of neck loading evaluated at two locations in the neck, upper and lower. The six modes of neck loading evaluated in the NIC are: 1) tension duration (+Fz), 2) compression duration (-*Fz*), 3) resultant shear duration (*Fx*, *Fy*), 4) Nij (composite of tension/compression (*Fz*) plus maximum instantaneous flexion/extension (*My*) as discussed previously), 5) maximum instantaneous lateral bending (*Mx*), and 6) maximum instantaneous twisting (*Mz*). In general, and where possible, the NIC limits correspond to a 10% risk of AIS 3+ neck injury, but this correlation is unclear; both the probability of injury and the injury levels are not clearly undergirded by robust risk functions. **Table 4** provides a summary of the 12 neck injury criteria, the formulation (if applicable), and the associated thresholds.

**Table 4. NIC Summary (from Nichols, 2006)**

<table>
<tr><th colspan="2"><u>Criteria Element</u></th><th><u>Upper Neck Limit</u></th><th><u>Lower Neck Limit</u></th></tr>
<tr><td colspan="2">1) Tension Duration<br>S – small (0-135 lb)<br>M – medium (136-199 lb)<br>L – large (200+ lb)</td><td>S (5 ms, 414 lbs<br>31 ms, 414 lbs<br>40 ms, 200 lbs<br>80 ms, 200 lbs)<br>M (5 ms, 618 lbs<br>35 ms, 618 lbs<br>45 ms, 320 lbs<br>80 ms, 320 lbs)<br>L (5 ms, 761 lbs<br>37 ms, 761 lbs<br>48 ms, 450 lbs<br>80 ms, 450 lbs)</td><td>Same</td></tr>
<tr><td colspan="2">2 ) Compression Duration<br>S – small (0-135 lb)<br>M – medium (136-199 lb)<br>L – large (200+ lb)</td><td>S (5 ms, 519 lbs<br>27 ms, 200 lbs<br>80 ms, 200 lbs)<br>M (5 ms, 790 lbs<br>30 ms, 320 lbs<br>80 ms, 320 lbs)<br>L (5 ms, 979 lbs<br>32 ms, 450 lbs<br>80 ms, 450 lbs)</td><td>Same</td></tr>
<tr><td colspan="2">3) Shear (composite) Duration<br>S – small (0-135 lb)<br>M – medium (136-199 lb)<br>L – large (200+ lb)</td><td>S (5 ms, 405 lbs<br>20 ms, 225 lbs<br>29 ms, 225 lbs<br>37 ms, 165 lbs<br>80 ms, 165 lbs)<br>M (5 ms, 625 lbs<br>25 ms, 337 lbs<br>35 ms, 337 lbs<br>45 ms, 247 lbs<br>80 ms, 247 lbs)<br>L (5 ms, 777 lbs<br>28 ms, 414 lbs<br>39 ms, 414 lbs<br>50 ms, 304 lbs<br>80 ms, 304 lbs)</td><td>S (5 ms, 810 lbs<br>20 ms, 450 lbs<br>29 ms, 450 lbs<br>37 ms, 330 lbs<br>80 ms, 330 lbs)<br>M (5 ms, 1250 lbs<br>25 ms, 674 lbs<br>35 ms, 674 lbs<br>45 ms, 494 lbs<br>80 ms, 494 lbs)<br>L (5 ms, 1554 lbs<br>28 ms, 828 lbs<br>39 ms, 828 lbs<br>50 ms, 608 lbs<br>80 ms, 608 lbs)</td></tr>
<tr><td>4)<br>$N_{ij} = \frac{F_z}{F_{zcrit}} + \frac{M_y}{M_{ycrit}}$</td><td><table><tr><th></th><th>S</th><th>M</th><th>L</th></tr><tr><td>$+F_{zcrit}$ (lb)</td><td>964</td><td>1530</td><td>1847</td></tr><tr><td>$-F_{zcrit}$ (lb)</td><td>872</td><td>1385</td><td>1673</td></tr><tr><td>$+M_{ycrit}$ (in-lb)</td><td>1372</td><td>2744</td><td>3673</td></tr><tr><td>$-M_{ycrit}$ (in-lb)</td><td>593</td><td>1195</td><td>1584</td></tr></table></td><td>Peak $Nij$ < 0.5</td><td>Peak $Nij$ < 1.5</td></tr>
<tr><td>5) $NMI_x = \frac{M_x}{M_{xLIM}}$</td><td><table><tr><td>$+/-M_{xLIM}$(in-lb)</td><td>593</td><td>1195</td><td>1584</td></tr></table></td><td>Peak $NMI_x$ < 0.5</td><td>Peak $NMI_x$ < 1.5</td></tr>
<tr><td>6) $NMI_z = \frac{M_z}{M_{zLIM}}$</td><td><table><tr><td>$+/-M_{zLIM}$ (in-lb)</td><td>593</td><td>1195</td><td>1584</td></tr></table></td><td>Peak $NMI_z$ < 0.5</td><td>Peak $NMI_z$ < 1.0</td></tr>
</table>

The NIC considers the set of ejection neck injury criteria as “success criteria” rather than black and white pass/fail criteria, due to the dynamic and complex nature of an ejection event (Nichols, 2006). The application of these criteria is described as a set of flags. If none of the criteria are failed during a test, then the test is a success with no caution flags raised. If one or more of the criteria are failed during a test, then a flag is raised and the issue is investigated to determine if it is truly pointing to a potential cause of injury (Nichols, 2006). This is accomplished by reviewing the details of the exceedence including body position, off axis neck loading, seat, chest, and head linear and angular acceleration, the portion of the limit curve that was exceeded, and the magnitude of the exceedence (Nichols, 2006). Depending on these details involved with an exceedence of one or more of the criteria in a test, the exceedence might be dismissed if it is considered low risk. On the other hand, it might be accepted if the details of occurrence support evidence that a neck injury hazard truly exists. The reader is referred to Nichols (2006) for further details including limit values for each of the six modes of neck loading for each anthropometric category (small female, mid-size male, and large male) as well as the equations and/or curves used to determine each criterion (Nichols, 2006). It should be noted, however, that the logic tree for evaluation of these exceedences is not documented.

The tension, compression, and shear force duration limits used in the NIC are based upon the Mertz automotive duration criteria discussed previously but have been modified for application to the ejection environment. According to Nichols, the short duration tension limits correspond to about a 10% risk of AIS 3 neck injury, and while the longer duration load limits also correspond to some injury mechanism, it is unspecified what this injury risk is in the NIC (Nichols, 2006). The reader is referred to the Nichols paper for detailed application of the duration limits to specific ejection neck load time history. The risk of injury for the compressive duration limits and the shear duration limits are also not specified or known. This presents one of the limitations of applying the duration limits in an effective neck injury criterion. The

duration limit curves depict a region where significant neck injury is unlikely and a region that represent potential for significant neck injury. What "significant" means and exactly what "potential" and "unlikely" mean are unknown. The vague and indeterminate nature of the duration curves used in the NIC make their use, and more specifically their justification, difficult in the application of the neck injury criteria to acceptance testing.

The NIC is like the Nij in that it provides a means for evaluating an ejection seat or component of the escape system based upon observed neck loads in the ATD. It provides an extremely limited risk prediction capability for the probability of various levels of AIS injury for the one element of the 12 sub-criteria for which a risk function has been developed, the upper neck Nij (though the validity of this risk function has been shown inadequate for the military aviation environment (Parr et al., 2013)). Other sub-criteria, which have only load limits but no risk curves, only afford binary injury prediction capability. The NIC is comprehensive in nature, incorporating multi-axial loading, which is experienced by the pilot in the ejection environment that would potentially cause harmful loading to the neck. It allows the safety of various systems being developed to be evaluated like an ejection seat modification, addition of an HMD, or a completely new aircraft escape system.

There are drawbacks of the NIC that are worth mentioning. While it sets limits on every potential pathway for injurious neck loading in the 12 elements of the criteria, some of these are redundant. It is possible for a test to pass the load duration tension or compression criterion but fail the tension or compression criterion embedded in the Nij. This redundancy makes the NIC difficult to use for making tradeoff analyses during system design. It also makes it difficult for program managers in the acquisition community to provide definitive requirements and specifications to their contractors. Once a prototype system is delivered for acceptance testing, it is then difficult to assess the system's adherence to those requirements during developmental testing. The combination of instantaneous peak and load duration values in the NIC make it a

fairly robust set of criteria, ensuring neck safety by protecting against both peak and load duration neck loading. However, as is the case in the NIC, the two are not in harmony and provide conflicting pass/fail or accept/reject values, this benefit is not realized. Additionally, the allowable risk in the criteria where risk curves are utilized is a 10% risk of AIS 3 or greater. The AFLCMC escape office's requirements of 5% risk of AIS 2 or greater are not met in the NIC. Also, NIC is built around data gathered from Hybrid III ATD on the sled track at Holloman AFB. Other researchers have critiqued the NIC and have suggested changes to improve the criteria relative to the conflicting standards that make for difficult system evaluation (Carter et al., 2000; Pellettiere et al., 2011; Pellettiere, 2012).

As an example, consider a single developmental ejection test of the F-35 escape system with a 50% male Hybrid III ATD as an example (see test summary in **Table 5**). This particular test was evaluated using the NIC. It failed the instantaneous Nij criteria portion of the test (combination of axial load and sagittal plane bending moment), but it passed the duration tension, compression, and shear limits. This provides conflicting results, since both criteria contain the same loading (tension and compression). This presents one of the major problems of a criterion with both duration and instantaneous neck load limits. Each criterion was constructed in a different manner with different underlying assumptions and supporting research and data. It would be extremely difficult to have criteria with both that did not result in mixed results.

Table 5. Summary Results from Upper Neck of ATD Rocket Sled Test

| Description | Maximum | Time of Maximum (Sec) | Minimum | Time of Minimum (Sec) | Risk | Limit |
|---|---|---|---|---|---|---|
| Fx Shear (Lb) | 207.89 | 0.1900 | -49.82 | 3.0190 | | |
| Tension (Lb) | 646.35 | 0.1890 | | | | |
| Compression (Lb) | -134.58 | 1.9210 | | | | |
| Flexion (In-lb) | 307.85 | 1.7410 | | | | |
| Extension (In-lb) | 283.22 | 0.2020 | | | | |
| Ntf | 0.1507 | 0.1630 | | | | 0.5 |
| Nte | **0.5578** | 0.1890 | | | **Exceeds** | 0.5 |
| Ncf | 0.1984 | 1.7430 | | | | 0.5 |
| Nce | 0.0596 | 7.1030 | | | | 0.5 |
| Composite Nij | **0.5578** | 0.1890 | | | **Exceeds** | 0.5 |
| Fy Shear (Lb) | 87.59 | 1.6100 | -76.36 | 1.6850 | | |
| Shear Resultant (Lb) | 214.06 | 0.1900 | 0.31 | 5.2450 | | |
| Mx (In-lb) | 166.80 | 2.2020 | -225.83 | 1.6850 | | |
| Mz (In-lb) | 70.92 | 1.7070 | -316.92 | 0.3820 | | |
| UNMIx | 0.18898 | 1.6850 | 0.00002 | 8.0450 | | 0.5 |
| UNMIz | 0.26521 | 0.3820 | 0.00000 | 7.9370 | | 0.5 |

While this specific combined set of criteria has some undesirable qualities, it might be possible with further research to include in a set of combined neck injury criteria a component with load duration in addition to a component with peak loading. This additional research would need to establish a way of translating duration limits into an injury risk assessment and further understand the causal link between load duration levels and neck injury. This would prevent the resulting criteria from conflicting with itself, being unnecessarily complicated for systems manufacturers to comply with, and being difficult to implement and assess system performance in developmental testing.

## FAA Neck Injury Criteria for Side-Facing Aircraft Seats

The Federal Aviation Agency (FAA) published Neck Injury Criteria for Side-Facing Aircraft Seats, the culmination of numerous government and academic research efforts to

understand human tolerance and injury thresholds in lateral inertial loading (FAA, 2011). The purpose of the research, according to the report, was to "investigate neck injuries in side-facing aircraft seats and to develop neck injury criteria and injury tolerance levels (Federal Aviation Administration, 2011: 69)." These criteria were developed using a matched pairing of whole body PMHS and ES-2 ATDs (an ATD designed as a human surrogate for evaluating side acceleration) (FAA, 2011). Test subjects (PMHS or instrumented ATD) were seated sideways in standard simulated rigid side-facing aircraft seats, restrained appropriately, and impacted at input levels specified by the FAA. The ATD provided researchers with the neck loading for a specific test condition while the PMHS (subjected to the identical test configuration) was evaluated to determine the injury level caused by the test. Throughout the course of the research and analysis, it was determined that peak instantaneous upper neck tension was significantly correlated to injury, thus the criteria were constructed around peak tension values observed in the ES-2 (FAA, 2011). This use of instantaneous peak tension as an injury predictor assumed, however, that either bending, shear, or torque loads were also present concurrently. It was noted that it is almost impossible in side acceleration for the neck to be subjected to pure axial tension, and thus it is assumed for the criteria that these other loads are present along with tension to cause injury.

As a result of the logistic regression performed on the data, it was determined that "the risk of serious neck injury for occupants of side facing seats can be limited to a 50% probability if the value of upper neck tension measured in an ES-2 ATD during seat qualification tests is $\leq$ 2300 N (Federal Aviation Administration, 2011: 73)." Additionally, for lower probability evaluations, it was determined from the logistic regression risk curve that an Injury Assessment Reference Value (IARV) of $\leq$ 1800 N of tension in the ES-2 ATD will limit AIS 3 + neck injury to less than 25% (FAA, 2011). This criterion serves as the basis for a "performance standard for the certification of side-facing aircraft seats and corresponding protection systems (Federal Aviation Administration, 2011: 69)." It should be noted that this criterion was built from logistic

regression of 10 PMHS accelerative tests. The report conceded that this was a minimal sample size for making any statistical claim. They used the probit method of logistic regression, which makes the assumption of a large sample size and exact data, neither of which hold for their data set, potentially adversely affecting the predictive capability of the risk function and the resulting injury criterion.

### USAF Interim HMD Criterion (aka the "Knox Box")

Perry and Buhrman summarized their work on the effect of helmet inertial properties on the biodynamics of the head and neck during +Gz accelerations and the resultant AF interim criterion used in the acquisition process for developing HMDs (Perry and Buhrman, 1996). The interim criterion (also known as the Knox Box – see **Figure 12** below) is stated as follows (applicable only to the catapult phase of ejection):

> "For ejection seats similar to the B-52 seat having a typical impact acceleration peak of approximately 18 G, helmets weighting less than 4.5 pounds and having a combined head/HMD CG within limits defined by -0.2 to 0.85 inches on the x-axis and 0.4 to 1.4 inches on the z-axis of the anatomical axis system of the head, will induce a risk of neck injury similar to current operational helmets. For ejection seats similar to the ACES II seat having a typical impact acceleration peak of approximately 12 G, helmets weighing less than 5 pounds and having CG limits of -0.2 to 1.1 inches on the x-axis and 0.4 to 1.4 inches on the z-axis of the anatomical axis system of the head, will also induce a risk of neck injury similar to current operational helmets (Perry and Buhrman, 1995:89)."

Since the Knox Box constrains the combined CG of the head and HMD, the criteria allows for a fairly wide range of HMD mass and CG combinations due to the larger magnitude of head mass relative to the HMD.

The paper also reports the experimental results of changes to varied helmet inertial properties on the biodynamic response of live test subjects under 10 Gs in the $+G_z$ acceleration. The maximum helmet weight of 7.5 lbs with a CG close to the anatomical y-axis resulted in a maximum compressive value of 260 lbs, which conforms to "previous studies citing maximum tolerable compression loads of 250 lbs without injury and below the threshold value cited for hard tissue injury (420 lbs) (Perry and Buhrman, 1996)." Shear and torque values with the heaviest helmet were below values cited for hard tissue thresholds for damage (437 lbs and 1700 in-lbs respectively) (Perry and Buhrman, 1996).

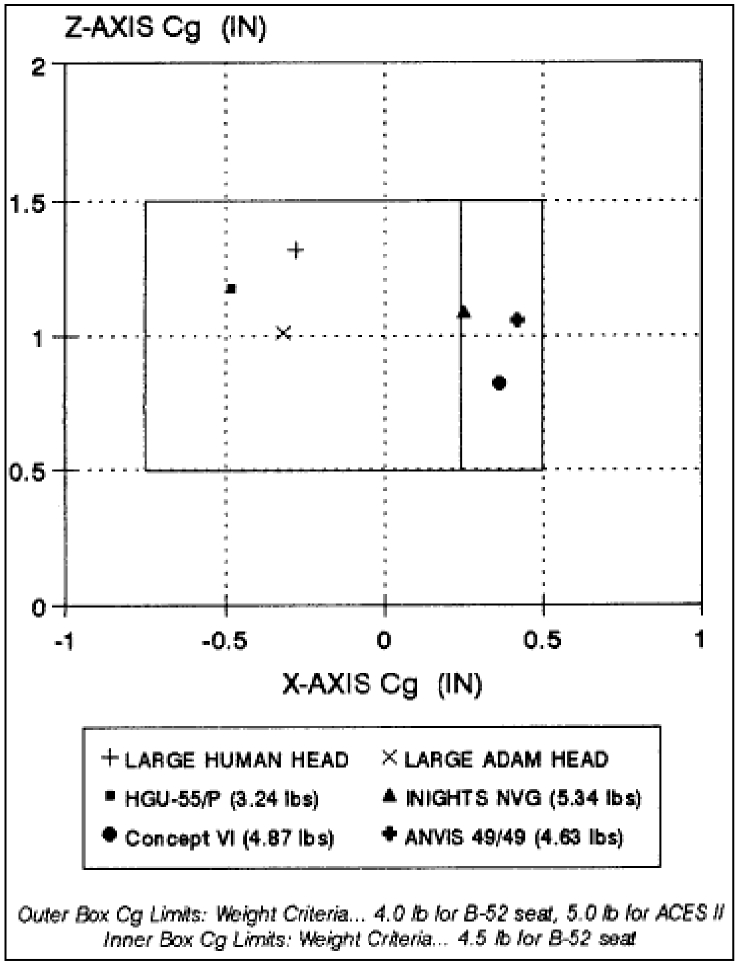


**Figure 12. Knox Box Plot** (Perry, 1994)

No injury risk information is explicitly defined in risk curves, and thus it makes decision making and safety tradeoff studies difficult based upon the parameters set forth in this criterion. It is also not clear what risk the occupant is exposed to by wearing a helmet outside of the box; all that is known is that it is greater than the risk associated with legacy systems. It is also only applicable to the catapult phase of ejection, thus it does not provide safety guidelines or injury risk prediction for Gx and Gy loading. The Knox Box was never intended to be the definitive or final Air Force design direction for HMDs. However, in the absence of other guidance or final

neck injury criteria applicable to conditions involving head supported mass, it has remained the standard used by program offices to guide the design of HMD mass properties.

### Tensile Neck Injury Criterion

Carter et al. developed a tensile neck injury criterion for use in the aviation ejection environment (Carter et al., 2000). Their research posits that other neck injury criteria were developed for application to environments where frontal flexion was the primary loading mechanism, as in the automotive industry for frontal crashes. Since frontal flexion is not the primary loading mechanism in aviation ejection, or many other environments, they sought an alternate criterion. In ejection, tensile forces act upon the pilot's neck during wind blast as well as parachute opening shock.

Logistic regression was used to generate a risk function for the data set, which consisted of 208 human non injurious data points and 10 PMHS injurious data points. Carter et al. were the first to generate injury risk curves based on a combined human/PMHS data set, a novel approach intended to better define injury risk curves at the lower force input values using human neck response data. The risk curves delineated between small individuals (under 73 kg) and large individuals (over 73 kg). The curves predict a 5% probability of AIS greater than 3 neck injury at tensile neck load of 2320 N for large individuals and 1740 N for small individuals, consistent with the findings of other studies (Carter et al., 2000).

Nusholtz et al. lend validity to focusing on tension only in their research, in which they conducted an analysis of the combined loading Nij criteria using previously published biomechanical data to determine if any of the neck loading pathways (tension, compression, flexion, or extension) predicted injury better than the others (Nusholtz et al., 2003). It was

observed that tension only identified the injury risk in frontal impact (-Gx acceleration). Neither compression, flexion, nor extension was able to predict whether injury would occur or not, and actually including these neck loads in the analysis decreased the criterion's accuracy of prediction (Nusholtz et al., 2003). The developers of the tensile neck injury criterion acknowledge that their criterion focused on unidirectional loading, which rarely occurs, and suggest that future research pursue understanding multi-axial loading as the next step for further enhancing this neck injury criterion (Carter et al., 2000). Chapter VI of this dissertation addresses the tensile neck criterion in depth and develops and proposes an updated AF tensile neck injury criterion.

## Criteria Analysis

Table 6 compares each of the criteria previously reviewed with the AFLCMC escape office's neck injury criteria requirements. Criteria that are able to account for occupant size have some means of assessing injury risk based on body mass either through use of critical values or providing separate curves for individuals of different body mass.

**Table 6. Existing Criteria Comparison Against AFLCMC Escape Office Requirements**

| | 5% Risk/AIS2+ Injury | Risk Function(s) | Multi-axial | Accounts for Occupant Size |
|---|---|---|---|---|
| **Mertz Criteria** | No | No | Partial | Yes |
| **Nij** | No | Yes | Partial | Yes |
| **Beam Criteria** | Yes | Yes | Partial | No |
| **NIC** | No | Partial | Yes | Yes |
| **FAA Side** | No | Yes | No | No |
| **Knox Box** | No | No | No | No |
| **Tensile** | No | Yes | No | Yes |

All of the criteria except the Mertz Criteria, the Knox Box, and the upper neck Nij subcomponent of the NIC have associated risk curves developed; this allows some flexibility for

decision makers to decide what level of risk is acceptable. The Nij and the similarly structured Beam Criterion considers four types of neck loading, specifically focusing on the combination of axial loading and anterior/posterior bending moment. The Nij is an upper neck criterion that was not designed for head supported mass (though it has been applied to assess neck loading with head supported mass(Parr et al., 2012; Nichols, 2006)), and the Beam Criterion was designed for the lower neck and does consider head supported mass. Like the Nij, the NIC also considers axial loading and bending (the NIC includes the upper and lower neck Nij as two of the 12 criteria). Axial loading is addressed twice in the NIC, first as the single force duration limit of neck tension and compression (both for upper and lower neck), and second as included in the Nij combination of axial loading and bending moment. Distinct from the all the other criteria, the NIC also considers shear, lateral bending, and neck twisting (all for both upper and lower neck). The NIC and the Beam Criterion are the only criteria that include lower neck loading, the latter of the two consisting of only lower neck loading with no consideration for upper neck loading. The FAA side criteria, the USAF Tensile neck injury criterion, and the Mertz Criteria are single force criteria. The Nij and the Beam Criterion are two-force criteria, but not fully multi-axial. The NIC is the only true multi-axial criteria of the group.

The Nij in its current automotive-specific form allows for a higher acceptable limit for injury risk than that acceptable for military aviation (22% risk of AIS $\geq$ 3 versus 5% AIS $\geq$ 2). It also is specifically designed for frontal crash. This is likely to be insufficient for ejection wherein other forces are present throughout the ejection sequence which might also be important from an injury mechanism standpoint. Additionally the Nij does not meet the AFLCMC requirement for a multi-axial neck injury criterion. Additional details of the shortfalls of the Nij

in the ejection environment are provided in Chapter IV. The NIC also allows for higher risk than the AFLCMC escape office is comfortable with (10% risk of AIS $\geq$ 3 at $N_{ij}$=0.5). The tensile neck injury criterion is more in line with AF risk acceptance (5% AIS $\geq$ 2), though it does not address the multi-axial requirement of the AFLCMC escape oversight office. The NIC is redundant in some of its constraints, which make it difficult to use as an evaluative tool. Refer to **Table 6** for complete detail of how each of these criteria fulfill the AFLCMC injury criteria requirements.

The Knox Box is specific to HMD mass properties. The fact that it is the only tool that provides some design guidance for the mass properties of HMDs accounts for its lasting popularity with the HMD acquisition community, despite it only being an "interim" criterion. The Knox Box has some shortcomings. The Knox Box gives designers very basic guidance for HMDs, imposing limitations on HMD CG based upon a maximum mass. It does not offer any optimization or tradeoff information to enable improved HMD design, nor does it involve risk curves to specify probability of neck injury based upon a given input. If the parameters of the HMD fit within the box, then it passes, since every point in the box has been tested safely on human subjects representative of the population of military aviators. While it offers quite a bit of leeway, permitting the center of gravity to be shifted 1.25" in horizontal and 1.0" in vertical zone, the lack of injury risk data makes it difficult to use as a criteria when making system level trades. It might be possible to find a suitable replacement or improve upon the guidelines provided by the Knox Box in a way that informs and guides better design and at the same time yields the production of a safe HMD that will not expose the pilot to undue neck injury in an ejection scenario.

Some work has been done to compare neck injury criteria in experiments with PMHS at injurious acceleration levels. Salzar et al. performed a study on three 5th percentile PMHS and seven 5th percentile ATDs to create a baseline body of data that investigates the ability of a 5th percentile ATD to predict whole-body kinematics observed in simulated aircraft ejection and whether current neck injury criteria are applicable (Salzar et al., 2009). In this study, the PMHS and ATDs were seated in a standard naval fighter ejection seat, restrained by a standard naval harness, and subjected to +Gz acceleration (mean peak acceleration of 22.9 G) (Salzar et al., 2009). The NIC, the Nij, and the Beam Criterion were evaluated to assess how each of the criteria performed in predicting injury for small individuals. They concluded that the Nij under predicted injuries observed in the PMHSs, the Beam Criterion over predicted, and the NIC best predicted injuries observed in the three PMHSs. The limitations of this study give some insight into these conclusions. The flexion critical intercept value used in the Beam Criterion was much lower than the Nij, possibly accounting for the over prediction of injury. Additionally, the Beam Criterion was developed using head/neck sections rigidly implanted at T1, which potentially translated the flexion into the specimen higher into the cervical spine, whereas in the PMHS tests the lower thoracic spine experiences the flexion (Salzar et al., 2009). Also, with only three data points, it is possible that the data collected were just a poor fit for the criteria tested; more data points might have changed the result of the comparison.

## Application of Field Data

There is a limited amount of real world accelerative neck response data available to researchers that contains detailed neck loads combined with precise injury classifications from

which to construct injury criteria or even enough data that would be required to validate injury models. Without the constraints of funding and invasiveness, it would be ideal to instrument every pilot in ejection seat equipped aircraft with an accelerometer to record head acceleration data during each mission, which can then be used to calculate neck loads. Unfortunately, cost and implementation problems have precluded observing neck loads in operational flying missions. The neck load data obtained from monitoring could be used to calculate neck loads experienced by the pilots either over a typical mission or in an ejection event if that were to occur. When paired with injury data from that ejection, this data could serve to perform a robust neck injury criteria model validation. The knowledge of typical mission neck loads would enable researchers to better understand the effects of head supported mass on neck fatigue, which is outside the scope of this study but a very important issue to program managers with the introduction of HMDs worn full time (e.g., F-35 and F-16/A-10 equipped with the Helmet Mounted Integrated Targeting system). The ability to access real world ejection neck load data would remove the guesswork around what neck loads pilots actually experience in an ejection.

As it stands today, the ATD neck load data from the rocket sled tests are the closest researchers get to being able to observe the neck loads experienced over the full sequence of ejection, but as stated previously the Hybrid III neck has been observed by some researchers in certain modes of loading to be an imperfect surrogate for human neck response with head supported mass (Buhrman and Perry, 1994; Bass et al., 2006; Salzar et al., 2009). Human neck loads in actual ejections can be estimated from rocket sled tests using an average of neck loads from ATDs ejected at similar speeds. These are only rough estimates, however, since the ATDs are not perfectly biofidelic and each ejection event is highly variable due to the dynamic nature of ejection.

NHTSA incorporated real world injury data to validate the Nij (Eppinger et al., 1999). They analyzed frontal collision data that included estimated collision speed, neck injury, and

whether the occupant was belted or not and estimated neck loads from corresponding ATD test frontal collisions to estimate neck loads. They concluded that their criteria performed in a satisfactory manner and that no adjustments to the injury risk curves were required (Eppinger et al., 1999).

The Air Force Safety Center and its counterpart in the Navy also keep records of real world ejection data, which is of some use to researchers since details of the ejection (i.e., airspeed, altitude, injuries sustained, etc.) can be used to correlate with ATD ejection sled test data to determine neck loading related to specific injury. Unfortunately, since these specific pilot neck load data are not collected, neck load can only be estimated, and therefore these correlated ATD data are of limited value.

**Load Duration Versus Peak Instantaneous Loading Criteria**

From the literature, there seems to be more support from a research and biomechanical perspective in favor of using instantaneous peak loading to develop future neck injury criteria (FAA, 2011; Eppinger et al., 2000; Parr et al., 2013). From the selected neck injury criteria assessed, neck injury criteria constructed using peak instantaneous loading provide more insight to understanding the link between load and injury compared with load duration criteria. If load duration alone is used, it would be hard to determine at which point injury occurs in the time duration. The load duration technique also does not lend itself to constructing risk functions, which have proven to be the key tool in establishing limits or IARVs based upon a known risk of injury. Limits derived from sound risk functions provide a clear-cut standard to be met by a system in acceptance testing.

Of the criteria reviewed in this chapter, only the Mertz criteria and a portion of the NIC that adapted the Mertz criteria for the ejection environment use a load duration criteria. The Mertz criteria were developed early in the car safety era, and in the current NHTSA Nij injury

criteria the duration limits have been removed (Eppinger et al., 1999; Eppinger et al., 2000). The duration values for the Mertz criterion are also a difficult performance standard to design to as discussed previously in reference to the confusion the duration limits add to the NIC. The vague injury risk prediction and lack of clear guidance that exist within the duration limits that often bring confusion is a possible reason for their removal from the Nij for evaluation of advanced restraint systems in automobiles. Perhaps the most significant argument for choosing instantaneous criteria is the fact that the researchers who developed the most recent neck injury criteria (Beam Criterion and FAA side-facing criteria) incorporated the instantaneous load method as the basis for their construct. The researchers who contributed to those criteria are experts in the field, some of the most respected and well published in biomechanics and injury prevention.

**Statistical Techniques for the Development of Injury Risk Functions**

This section will provide a review of pertinent methods of statistical modeling used to produce injury risk curves. Injury risk curves allow decision makers to design and evaluate systems to a specific level of acceptable risk and serve as the foundation of any injury criterion (Pellettiere, 2012). These curves are formed using various statistical techniques that model injury probability as a function of some input (Cutcliffe et al., 2012). Further, these models define the risk of injury based upon analysis of experimental data with either specific force input or a combination of forces for input and a pre-specified binary outcome (injury/no injury) as the dependent variable. The first method discussed is logistic regression, which has been widely used to produce injury risk curves in previous work. The second method is survival analysis (SA), which is growing in acceptance and frequency of use amongst researchers in this field based upon the way it uniquely handles the specific characteristics of human and PMHS data. An overview of each method is covered in the next two sections.

### Logistic Regression

Logistic regression (LR) is commonly used in data analysis where researchers desire to model an association between a binary or dichotomous response variable and one or more predictor variable(s) (Hosmer and Lemeshow, 2000). As such, it may be natural to model an injury risk curve using injury/no injury as a response. Indeed, logistic regression has been used in the literature in the past to generate injury risk functions (Eppinger et al., 1999; FAA, 2011; Carter et al., 2000).

However, LR has limitations. Importantly, LR assumes each data point is exact. In the field of biomechanics, data is often gathered in such a way that the exact value of an observed neck loading result is unknown. In the case of injurious testing, the actual value of the loading that caused the injury may be less than the loading value recorded (Cutcliffe et al., 2012). This type of data is referred to as being left-censored. On the other hand, in the case of non-injurious, human subject testing, the actual value of the loading that might cause injury is greater than the loading value recorded. This type of data is referred to as being right-censored. An injury risk curve that seeks to incorporate both human and PMHS data would be using both left and right censored data. Logistic regression also assumes a large sample size ($N > 100$), which is typically not feasible in both human subject and PMHS experiments. One possible solution is to use Firth's adjusted maximum likelihood method of LR. This method may be used as a correction, where LR coefficients might be biased when data is skewed toward one outcome (injury or no injury). Therefore, Firth's adjusted maximum likelihood method may be appropriately applied when using data that is either left- or right-censored.

Firth's method is also useful when there is a small sample size ($N \leq 100$) or when the contingency table (for discrete predictors and outcomes) has too many cells with low counts (Firth, 1993). Due to the expense of collecting injurious data points, this attribute of Firth's maximum likelihood is also very desirable. Although this method may be used to make the LR

model more appropriate for such data, methods which are applicable to small sample sizes and that are capable of handling data which may be both left- and right-censored are desired.

### Survival Analysis

Survival analysis (SA) has recently been applied in the field of injury biomechanics to generate injury risk functions (Hosmer et al., 2008; Cutcliffe et al., 2012; Bass et al., 2006). Adoption of this technique is partially due to advances in computing capability and the incorporation of survival analysis techniques in “point and click” statistical software. Furthermore, SA supports the use of censored data and can support data sets where a portion of the data is left-censored while another portion is right-censored. Sample size is not restricted in applying SA to a data set as it is with LR. In order for parameters to be estimated using either SA or LR, the data must have at least one overlapping point in the data between injury and non-injury.

If all of the injurious PMHS data points have a higher criteria value than all of the non-injurious PMHS data points (data separation), regression using SA is not possible as the method does not converge on a solution. If the observed result of the PMHS accelerative experiments is complete separation of the data, researchers must use a less desirable statistical technique to generate a risk function that may not account for censoring in the data. In some cases, it is possible that complete separation in the data occurs at one AIS level, which might result in the ability to create a risk function at certain AIS levels but not others. This also has the potential to limit the analysis researchers can perform. It is possible for the value of a single PMHS data point to make or break the process of constructing risk functions.

## Summary

This section provided the background and reviewed the literature pertinent to the research objectives of this work. This body of current knowledge will serve as the foundation and basis for the subsequent chapters. The next chapter details the methods followed to construct the risk functions that comprise the improved set of neck injury criteria.

# III. Risk Function Construction Methodology

## Overview

This chapter describes the research methods followed to accomplish the objectives set forth in Chapter I related to risk function construction. This methodology is applied to construct the Gx, Gy, and Gz risk functions in Chapters IV, V, and VI. First, the fundamental elements of the criteria are established. Then, the methods used to develop the risk functions are detailed. Finally, the method for applying the criteria to quantify risk is described. Risk quantification is a key element that is required in order to outline HMD capability versus safety in the HMD HSI trade space.

## Criteria Fundamentals

Based upon the review of the literature discussed in Chapter II, the following fundamental elements should be incorporated into improved AF ejection neck injury criteria. First, criteria must clearly communicate the risk of injury and define the specific injury level associated with the risk. This is achieved by performing a survival analysis injury risk model constructed from sound biomechanical experimental data from human subjects, PMHS, or ATDs. Risk functions allow the decision makers to determine the appropriate level of risk that is acceptable, which will then determine the specific limit for the acceptance testing. A test of a system that exceeds the limit fails the criteria; otherwise the system is acceptable. Second, improved aviation neck injury criteria should present a consistent limit for the neck loads being evaluated. Third, the criteria should be based upon peak instantaneous loading rather than load duration. Finally, the improved criteria should satisfy as many of the AFLCMC escape office's neck injury criteria requirements as possible. The AFLCMC escape office has requested that future neck injury criteria should address the following requirements: 1) minimize the number of

criteria to simplify the determination of an acceptable escape system test, 2) be multi-axial, 3) account for head supported mass, 4) account for the full range of the expanded pilot population (103 to 245 lb), and 5) be clearly tied to injury risk such that an acceptable injury rate is a 5% risk of AIS 2 or greater neck injury. The 5% injury rate is a requirement for any single portion of the pilot population. For example, lower probabilities of injury observed in large males cannot be traded for higher probabilities of injury observed in small females. The 5% injury rate is also a requirement that should be met across the range of relevant airspeeds.

### Structure and Data Requirements

It was desired for the structure of the multi-axial neck injury (MANIC) criteria to include all six major forces that could be observed in the upper neck as a result of accelerative loading to meet the AFLCMC escape office multi-axial requirement. The form of the MANIC is proposed in Equation 2, based upon preliminary work done in this area by Perry et al. (Perry et al., 1997):

$$MANIC = \sqrt{\left(\frac{Fx}{Fxcrit}\right)^2 + \left(\frac{Fy}{Fycrit}\right)^2 + \left(\frac{Fz}{Fzcrit}\right)^2 + \left(\frac{Mx}{Mxcrit}\right)^2 + \left(\frac{My}{Mycrit}\right)^2 + \left(\frac{Mz}{Mzcrit}\right)^2} \quad (2)$$

The numerators of each term in this equation are each of the six major upper neck loads an occupant could be exposed to as recorded in the experiment. The denominators are the critical values established for each type of loading based upon subject body weight from the literature. The square root of the sum of squares formulation removes any negative numbers (e.g. –*Fz* is compression but only the magnitude of the load is of concern) and allows for the response to be dominated by relatively larger values in the input variables which serves the desired purpose to capture the important neck load responses in each axis of acceleration. If alternate formulations are found to be more appropriate, such as the pure sum of the absolute value of loads, that structure will be described in the appropriate chapter. Critical values are used successfully in the

NIC and the Nij and will be incorporated into the set of improved criteria. The critical load values (also called intercept values) perform two important functions in the set of criteria. First, they assign relative importance to each mode of loading in the MANIC equation based upon observed biomechanical properties of the neck relative to injurious pathways; that is, they normalize each of the loads and moments based upon the injury threshold of each individual load and moment component. Second, they allow the set of criteria to be normalized to occupant size, as well as to a desired numerical value for ease of use. As such, anthropometric differences relative to body mass are accounted for within the set of criteria.

In each axis of acceleration, some forces were more predominant, but an attempt was made to account for all forces in the criteria as much as existing data would allow. During the course of the research, it was discovered that some data sets were lacking one or more of these measured loads, thus a reduced form of the model was used based upon axis specific data availability. Chapter VII summarizes the data availability for each axis and provides the resulting structure of the MANIC(Gx), MANIC(Gy), and MANIC(Gz) subcomponents. The dominant forms of neck loading in each axis of acceleration that were available from the existing data were incorporated into the set of criteria and contain adequate data for a pilot-scale injury prediction model.

The availability of adequate data was important in this work. The risk functions produced in this work were constructed using combined human and PMHS data sets in order for human tolerance to loads to be accurately characterized in the resulting injury risk functions and resulting criteria. For statistical integrity, where possible, data from a single human subject experimental setup was paired with data from a single PMHS experimental setup. When it was necessary to combine multiple PMHS neck load and injury data sets to achieve a reasonable sample size, appropriate statistical tests on the mean were conducted to ensure combining the data was appropriate. The pairs of combined human/PMHS tests for each axis of acceleration

came from experimental set ups that were as similar to each other as possible and that contained the greatest sample size available. The human subject sample sizes were larger than the PMHS sample sizes due to the expense and difficulty of performing experiments with PMHS and the resulting paucity of data in the literature. To be able to calculate the full MANIC, each data set was required to have complete time history upper neck (OC) load data that included *Fx, Fy, Fz, Mx, My,* and *Mz*. Where data sets were lacking, a reduced form of the model was adopted as described in Chapter VII. These reduced forms of the model in each axis of acceleration still incorporate the dominant forces for that specific axis of acceleration. As such, the resulting combined criteria, which are made up of the three axis-specific sub-criteria from -Gx, Gy, and -Gz, are still considered multi-axial since the combined criteria include all six primary loads. Additionally, to construct the risk functions, the data for each PMHS subject was required to include the defined level of injury observed using the AIS scale, which provides clinical definitions along with a rating from zero to six of injury severity (AAM, 2008).

### Statistical Methods

Based upon the literature's documentation of the departure from LR and movement toward SA in the biomechanical injury risk field, the statistical method used to construct the risk curves in each axis of acceleration was SA, performed in Minitab statistical software (Version 16). The human subject data is right-censored and will be treated as such in the survival analysis, and the PMHS data is left-censored. Nonparametric tests on the mean were necessary to compare various groups and entities; this was done using the SPSS statistical software package (Version 18). Statistical trends and significant differences or similarities were noted and discussed.

## Risk Functions Development

Developing the risk functions primarily followed work done by Pellettiere (Pellettiere, 2012); his method was applied to formulate injury criteria. The steps are: 1) identifying the injuries, 2) defining the environment, 3) specifying the input energy, 4) conducting specific testing to generate injuries, 5) performing regression analysis, and 6) developing test procedures (applying the criteria) (Pellettiere, 2012). The steps below were applied to develop three separate risk functions for each primary axis of acceleration (Gx, Gy, and Gz). The goal is for the risk functions for each axis of acceleration to incorporate as many of the six primary neck loads as possible.

### 1) Defining the Injuries

The first step in the Pellettiere methodology is defining the injury level of concern. Neck injuries to ligaments, vertebrae, or the spinal cord due to inertial loading of the head and neck from the dynamic ejection environment are of primary concern for the criteria. It was assumed these injuries were a result of loads observed at the upper neck (OC). This work primarily incorporated the AFLCMC escape community requirements for injury criteria developed to a 5% risk of AIS 2 or greater neck injury, though AIS 3 or greater risk functions were developed as well for the sake of comparative analysis with legacy criteria. Chronic neck injuries and fatigue were not considered in the criteria, only acute injury resulting from the accelerative forces of ejection.

### 2) Defining the Environment

The second step in the methodology is defining the environment. For the purpose of this study, the environment is defined as the aviation ejection environment, where the occupant is

seated in an ejection seat, restrained in typical seat restraints, and exposed to the forces described previously during the complete ejection sequence. Specifically for the development of laboratory based risk curves for each axis of acceleration, the environment is defined as frontal acceleration (-Gx), side acceleration (Gy), and vertical acceleration (-Gz) to the seated pilot.

### 3) Energy Input

The third step is defining the energy input. The data used in this study came from experiments where energy input included a range of injurious and non injurious accelerative forces from 4 to 10 Gs in human studies and from 8 to 40 Gs in PMHS and ATD studies. Additionally, subject anthropometry affected tolerance to energy input and was accounted for in the criteria using critical values for each mode of loading (similar to those used in the NIC and Nij) (Pellettiere, 2012). The individual axis criteria chapters (Chapter IV, V, and VI) provide detail as to how each sub-criterion of the MANIC incorporated critical values into the formulation.

### 4) Specific Testing

Three different risk curves were generated using the previously performed experimental results of testing from the three major axes of acceleration input. Specific testing of humans and PMHS in -Gx, Gy, and -Gz accelerative input resulted in three separate data sets, which were used to create three separate risk functions. For this work, no new human or PMHS experiments were performed. Ideally, for risk functions based upon human neck response as this study seeks to construct, data on human subjects and PMHS would come from the same experimental set up to control as much variability as possible. Unfortunately, no experiments have been performed in this manner with the required load and injury data. Therefore, human and PMHS data from

studies with as much similarity as possible were sought. Typically, PMHS experiments have small sample sizes due to the expense and delicate nature of testing these types of subjects. Data from human subjects representing the military population (young and in good physical condition) were chosen where possible. There was a limitation in this area when it comes to PMHS data. Typically PMHSs are older and in poorer physical condition. Existing human subject data was culled from the AFRL Biodynamics database, and existing PMHS data was gathered from the literature or acquired through governmental/academic interagency partnerships.

**5) Regression Analysis**

Survival analysis was the primary statistical tool used to create the injury risk curves. The tensile (-Gz) criterion developed in Chapter VI incorporates LR and Firth's Method of LR to explore the differences in the risk functions produced by each statistical tool. This step in the methodology fulfills the AFLCMC escape office requirements that neck injury criteria be tied clearly to a defined probability of injury. Predicted values at significant injury risk percentages were also compared. This was done primarily at the 5% risk of injury level based upon the guidance from the AFLCMC escape office, though the risk functions afford flexibility to assess any risk level desired. The MANIC value predicted by the 5% risk is considered a preliminary, pilot-scale limit value. Assuming the risk curve is statistically robust and the decision makers have concluded that 5% is the determined acceptable risk of injury, the MANIC value associated with 5% risk of AIS 2 or greater injury would be the metric for future testing.

The previous steps (1 to 5) were applied to each axis of acceleration to develop three separate risk curves. This work is outlined in subsequent chapters. Chapter IV outlines the development of the -Gx (frontal acceleration) risk function, Chapter V outlines the development of the Gy (side acceleration) risk function, and Chapter VI outlines the development of the -Gz

(tensile load) risk function. Chapter VII brings the three sub-criteria together and describes the application of the complete MANIC to real world ejection data sets.

### 6) Test Procedures (Applying the Criteria)

*Individual Axis Criterion Application:* In this step, the individual, axis-specific risk function generated as a result of the preceding steps is used to evaluate the acceptability of a system or to perform a trade-off analysis on safety systems (such as head restraints) being used to mitigate neck injury risk. The test procedures to apply the criterion must be similar to those used in creating the criterion. In this case, the risk functions were applied to human, real world ejection, and PMHS studies data not used to construct the risk functions. For example, the risk curve developed for the Gx axis of acceleration was applied to predict the risk of neck injury from Gx human neck response data from a 10 G / 1.4 kg HMD experiment and compare it to a 10 G / 0 kg HMD experiment in Chapter IX. Thus, injury risk from added head supported mass can begin to be understood. This type of quantitative risk analysis was incorporated into a broader qualitative HSI examination of HMD capability versus safety discussed in Chapter IX.

*Combined Criteria Application:* The combined criteria were applied to existing ATD neck loads from real-world escape system qualification rocket sled testing in Chapter VII. This was done to demonstrate the feasibility of applying the criteria and to preliminarily assess the criteria's performance against legacy criteria. The assumption was made that the observed ATD neck loads are approximately similar to human neck loads. First, the MANIC value corresponding to the 5% injury risk for each of the individual curves was determined, and set as the limit for that specific mode of acceleration. Then, the ATD upper neck loads from the rocket sled tests were used to compute MANIC(Gx), MANIC(Gy), and MANIC(Gz) time histories for the full ejection sequence. Finally, the MANIC(Gx), MANIC(Gy), and MANIC(Gz) limits for each axis of acceleration were compared to the peak MANIC values observed in the ATD time

histories. If the observed ATD loads exceeded the MANIC(Gx), MANIC(Gy), or MANIC(Gz) limit, then that portion of the ejection test failed to meet the criteria. Additionally, due to the fact that the risk functions developed for Gx, Gy, and Gz acceleration are predictive of a specific percentage risk of injury given a load, the risk functions provide quantifiable risk posed by the condition tested rather than just a pass/fail assessment.

In order for the combined three-axis pilot-scale MANIC to be fully ready for implementation into a final USAF qualification standard to evaluate the safety of an escape system or HMD in developmental testing using an ATD, scaling would be required to carefully match ATD neck loads to human neck loads, as ATDs are not perfectly biofidelic. This scaling is outside the scope of this work and is recommended for follow-on research. This present research serves as a basis upon which to build the final developmental testing criteria.

# IV. Neck Injury Criteria Formulation and Injury Risk Curves for the Ejection Environment: A Pilot Study

## Chapter Overview

The paper that comprises this chapter has been published in the Journal of Aviation, Space, and Environmental Medicine (Parr et al., 2013). In this paper, the risk function development methodology was applied to a frontal acceleration (-Gx, see **Figure 13**) human subject and PMHS upper neck data set. The Nij formulation of neck loading was used to develop risk curves appropriate for the aviation environment using the survival analysis method of regression. Additionally, injury risk curve development methods are discussed. This study demonstrates the implementation of the methodology outlined in Chapter III to an existing legacy neck injury criterion and lays the framework for the application of these methods to create the MANIC(Gx) element of the multi-axial neck injury criteria developed in the subsequent chapters of this dissertation.

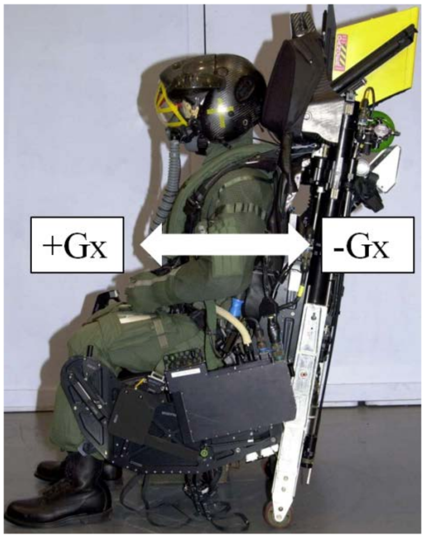


**Figure 13. Gx Axis of Acceleration**

**Abstract**

**BACKGROUND:** Helmet Mounted Displays provide increased pilot capability but can also increase the risk of injury during ejection. NHTSA's $N_{ij}$ metric is evaluated for understanding the impact of helmet mass on the risk of injury and modified risk curves are developed which are compatible with the needs of the aviation community. **METHODS:** Existent human subject data collected under various accelerative and head loading conditions were applied to understand the sensitivity of the $N_{ij}$ construct to changes in acceleration and helmet mass, as well as its stability with respect to gender, body mass, neck circumference, and sitting height. A portion of this data was combined with data from an earlier post-mortem human subject study to create pilot study modified risk curves. These curves are compared and contrasted with the NHTSA risk curves. **RESULTS:** A statistically significant difference in the peak mean $N_{ij}$ was observed when seat acceleration increased by 2 Gs but not when helmet mass was varied from 1.6 kg to 2 kg at a constant seat acceleration of 8 Gs. Although NHTSA risk curves predict a 13% risk of AIS2+ injury for the 8 G, 2 kg helmet condition mean $N_{ij}$ of 0.138, no AIS2+ injuries were observed. Modified risk curves were produced which predict a 0.91% risk of AIS2+ injury under these conditions. **DISCUSSION:** The $N_{ij}$ was shown to be sensitive to changes in acceleration, and generally robust to anthropometric differences between individuals. Modified risk curves are proposed which improve risk prediction at lower $N_{ij}$ values.

**Keywords:** HMD; pilot; aviation safety; risk curves

## Introduction

Helmet Mounted Displays (HMDs) are becoming human-machine interface equipment in manned flight. They have been shown to increase the performance of operators in their weapon system and thus increase overall mission effectiveness by adding capabilities such as enhanced night vision, and information fusion, which have the potential to enhance mission effectiveness across the spectrum of military operations (Rash et al., 2009). Unfortunately, this increased capability is often accompanied by increased mass, which can threaten pilot safety during ejection (Lewis, 2006; Nakamura, 2007; Stemper et al., 2003) and contribute to chronic neck and back injuries (Coakwell et al., 2004; Melzer, 2001). Of particular interest, the increased mass has the potential to increase the risk of operator neck injury if the pilot is subjected to accelerative environments like ejection. Injury due to a heavier HMD in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures and ligament ruptures (Buhrman and Perry, 1994; Stemper et al., 2003). Pilot anthropometric factors may also affect the likelihood of injury from neck loads induced by head-born mass; and recent changes in DoD manning requirements have increased the diversity of anthropometric characteristics among pilots (Harris, 1997). Therefore, it is important that pilot neck response be understood and characterized using a standard evaluation criteria that considers the influence of pilot anthropometric and biomechanical characteristics.

The National Highway Transportation Safety Administration (NHTSA) has established a frontal impact neck injury criteria ($N_{ij}$) for assessing risk of severe injury in automotive crashes (Eppinger et al., 1999; Eppinger et al., 2000). This criteria provides a quantitative method for evaluating and differentiating automotive crash and restraint systems where the quantitative metric can be related to the likelihood of injury in specified severity

categories. This metric has a strong foundation in biomechanics and relies upon results of crash tests with standardized Hybrid-III Anthropomorphic Test Devices (ATDs) to provide criteria for predicting the likelihood of injury to persons with varying anthropometric characteristics (Eppinger et al., 1999). The ability to define a relationship between the performance of the automotive crash and restraint system and the likelihood of injury is a key attribute of the $N_{ij}$ criteria, which does not exist for any known HMD or escape system evaluation method.

This research seeks to understand the applicability of the $N_{ij}$ formulation, or a more comprehensive criterion having similar characteristics, to the evaluation of helmet systems of varying mass in an accelerative aviation environment. Specifically, this research employed archived, Air Force (AF) frontal impact ($-G_X$) data (Doczy et al., 2004) to address the following questions:

1) Is the $N_{ij}$ formulation sensitive to changes in acceleration and helmet mass?
2) Is the $N_{ij}$ formulation sensitive to variation in anthropometric characteristics; including gender, body mass, neck circumference, and sitting height for subjects who are exposed to variations in acceleration and helmet mass?
3) Are the NHTSA neck injury risk curves applicable to the aviation accelerative environment and, if not, what is an appropriate family of risk curves?

NHTSA's neck injury criteria, the $N_{ij}$, established critical limits in four types of neck loading which are dominant in frontal impact automotive crashes involving accelerative forces primarily in the $-G_X$ axis. This criteria includes axial loading ($F_Z$, tension and compression), and sagittal plane bending moments ($M_Y$, flexion – forward, and extension – rearward) using a methodology initially presented by Klinich et al. (Klinich et al., 1996). Development of this injury criterion included applying previous biomechanical neck load and resultant injury research involving volunteer humans, porcine subjects and post-mortem human subjects (PMHSs). This

same research established critical limits for these four load pathways (Eppinger et al., 2000). The formula used to calculate the $N_{ij}$ is;

$$N_{ij} = \frac{F_Z}{F_{\text{int}}} + \frac{M_Y}{M_{\text{int}}} \tag{3}$$

In this equation, $F_Z$ and $M_Y$ are specifically observed instantaneous peak upper neck loads in a test automotive crash with the appropriately sized Hybrid-III ATD (i.e. small sized female, midsized male, and large sized male) designed to evaluate the performance of a restraint system. The values of $F_Z$ and $M_Y$ are the simultaneous instantaneous peak values that result in the largest $N_{ij}$ over the time history of the test. The values $F_{Zint}$ and $M_{Yint}$ are critical load values established by NHTSA for the maximum axial load in tension or compression and the measured flexion/extension bending moment established for various occupant size ATDs (Eppinger et al., 2000). The critical load values (also called intercept values) perform two important functions in the criteria. First, they assign relative importance to each mode of loading in the combined-force $N_{ij}$ equation based upon observed biomechanical properties of the neck relative to injurious pathways (e.g., they normalize the axial load and the bending moments based upon the likelihood of these individual components to induce injury). Second, they allow the criteria to be normalized to occupant size, as well as, to a desired numerical value for ease of use. As such, anthropometric differences are accounted for within the criteria.

Injury risk curves allow decision makers to design systems to a specific level of acceptable risk and serve as the foundation of any injury criterion (Pellettiere, 2012). These curves are formed using various statistical techniques, most commonly logistic regression or survival analysis, modeling injury probability as a function of some input, in the current case neck loading in the form of the $N_{ij}$ (Bass et al., 2006; Cutcliffe et al., 2012). These models define

the risk of injury based upon statistical analysis of experimental data with specific force input, or combination of forces, resulting in a binary outcome (injury/no injury) as the dependent variable specified at a certain defined injury level.  Risk curves were generated for the NHTSA $N_{ij}$ based upon a logistic regression of paired porcine injury and ATD neck load data which were scaled to develop limits for acceptable risk of injury to human occupants (Eppinger et al., 1999).  Specific injury level for each curve is based on Abbreviated Injury Scale (AIS) classification (AAM, 2008).  Based upon the consensus that no more than a 22% risk of AIS 3 or greater neck injury was acceptable, NHTSA applied the AIS 3 curve to select $N_{ij}$=1.0 as the performance limit (Eppinger et al., 1999).  Similar risk curves were constructed from the data for AIS 2, 4, 5, and 6 injuries.  Within the automotive application, a test that produces Hybrid-III ATD neck loads that exceed a $N_{ij}$ value of 1.0 fails the criteria.

When a pilot ejects from an aircraft he or she is subjected to four different phases, each phase exposing the pilot to different forces.  In order, these phases are: catapult stroke, windblast, seat stabilization, and parachute opening shock.  Most aviation-specific ejection studies have focused on the effects of the first phase, catapult stroke, in which the accelerative forces presented by the ejection mechanism act upon the head and neck in the positive z axis (upward, or $+G_z$).  However, the accelerative forces during all phases of ejection are a concern with increased helmet mass.  These additional phases can result in accelerative forces acting in the other major planes, $-G_X$ and $G_Y$, respectively.  An aviation specific neck injury criterion may need to consider each of these forces.  The current study focuses on forces in the $-G_X$ plane, consistent with the windblast phase of the ejection sequence, as this phase can provide forces similar to those experienced during frontal impact and permits the application of the NHTSA $N_{ij}$ neck formula to existing human data.  Research within the aircraft community has demonstrated that compressive and shear neck load, as well as neck bending moments, typically increased

linearly with increases in acceleration and helmet mass (Buhrman and Perry, 1994). Further studies have investigated the effect of helmet mass in accelerative environments within the other major axes ($G_X$ and $G_Y$) and compared male and female subjects in impact tests to expand the field of knowledge relevant to the smaller individuals and to ensure this population was not put at undo risk as a result of heavier HMDs (Buhrman and Mosher, 1999; Buhrman et al., 2000; Buhrman and Wilson, 2003). Although, early studies recommended that total helmet mass should be kept under 2 kg to prevent injury to pilots (Buhrman and Perry, 1994), a more comprehensive criterion, analogous to the $N_{ij}$, has not been developed within the aviation community.

The use of the $N_{ij}$ has been proposed as part of an overall neck injury criteria to evaluate aircraft escape system safety using ATDs as human surrogates (Nichols, 2006; Pellettiere, 2012). However, to the authors' knowledge this criterion has not been evaluated, qualified, or verified using human neck response data as an evaluative tool for HMD and escape system design. Within this application an $N_{ij}$ performance limit of 0.5 which corresponds with a 9.6% risk of AIS 3 or greater has been proposed, rather than NHTSA's 1.0 limit (AAM, 2008; Nichols, 2006). The lower performance limit was selected because a military pilot must be capable of avoiding capture or navigating to an extraction point after ejection, while NHTSA requires that a passenger survive an accident under the assumption that first responders will arrive on site to attend to any injuries. The escape system oversight office of the Air Force Life Cycle Management Center (AFLCMC), has clarified the requirement for AF aviation, specifying that a neck injury criteria be developed to evaluate HMDs and new escape systems such that acceptable injury rate should be 5% at an AIS 2 (moderate injury) (White JE. Personal communication; 2012). In addition to the need for a comprehensive criterion, further development of the injury risk curves to meet AFLCMC requirements is also required.

## Methods

Data from a previously performed human subject experiment on the effects of variable helmet mass on neck response to $-G_X$ acceleration (Doczy et al., 2004), which might represent the acceleration sustained from a frontal automotive impact or parachute opening phase of ejection, was used to understand the effects of interest on $N_{ij}$ response. The test "HMD" was a standard AF flight helmet (HGU-55/P) modified to allow variable mass to be attached to the helmet, which was properly fitted and attached to the subject's head using standard chin straps. For ease of reference this test helmet will subsequently be referred to as the HMD.

*Subjects*

Data from three experimental test configurations were used in this analysis. In the first, 26 human subjects wearing a 2 kg HMD were subjected to 6 Gs of accelerative force. In the second, 24 subjects wore a 1.6 kg HMD and were subjected to 8 Gs, and in the third, 23 subjects wore a 2 kg HMD and were exposed to 8 Gs. Detailed information for the specific subjects participating in each of the tests is shown in Table 7.

**Table 7. Human Subject Anthropometry and Peak Instantaneous Upper Neck Loads**

| | Subject Anthropometry | | | | | | Test Conditions: 8 G, 2 kg HMD | | | Test Conditions: 8 G, 1.6 kg HMD | | | Test Conditions: 6 G, 2 kg HMD | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Body Mass (kg) | Gender | Ht (cm) | Sit Ht (cm) | Age | Neck Circ | My (N-m) | Fz (N) | Nij | My (N-m) | Fz (N) | Nij | My (N-m) | Fz (N) | Nij |
| | 54.9 | F | 154.9 | 81.9 | 20 | 32.5 | - | - | - | 34.4 | 182.7 | 0.26 | 24.8 | 2.7 | 0.16 |
| | 60.8 | F | 158.8 | 84.3 | 29 | 31.2 | 46.6 | 152.7 | 0.17 | 24.8 | 153.0 | 0.10 | 15.4 | 115.2 | 0.07 |
| | 65.3 | F | 160.0 | 88.3 | 28 | 32.9 | 44.6 | 192.0 | 0.17 | 43.6 | 321.6 | 0.19 | 23.9 | 61.5 | 0.09 |
| | 65.8 | F | 175.3 | 91.4 | 19 | 31.5 | 32.8 | 230.3 | 0.14 | 32.6 | 203.7 | 0.14 | 48.4 | 41.9 | 0.16 |
| | 66.2 | M | 172.7 | 91.4 | 24 | 36.9 | - | - | - | 27.1 | 240.1 | 0.12 | 19.5 | 45.7 | 0.07 |
| | 68.0 | F | 165.1 | 84.5 | 27 | 33.0 | - | - | - | 40.6 | 15.4 | 0.13 | 25.2 | 188.4 | 0.11 |
| | 69.9 | F | 165.1 | 88.3 | 46 | 31.9 | 32.9 | 126.3 | 0.12 | 39.3 | 470.7 | 0.20 | 28.0 | 32.0 | 0.10 |
| | 72.6 | F | 170.2 | 88.9 | 23 | 35.8 | 40.5 | 111.2 | 0.15 | 34.7 | 122.8 | 0.13 | 24.8 | 12.6 | 0.08 |
| | 73.5 | F | 167.6 | 87.6 | 28 | 33.1 | 39.5 | 379.1 | 0.18 | 31.2 | 183.6 | 0.13 | - | - | - |
| | 73.5 | M | 180.3 | 94.0 | 35 | 36.7 | - | - | - | - | - | - | 18.1 | 3.0 | 0.06 |
| | 73.9 | F | 175.3 | 95.3 | 25 | 32.2 | 19.9 | 103.5 | 0.08 | 23.5 | 103.0 | 0.09 | - | - | - |
| | 73.9 | M | 188.0 | 100.3 | 30 | 35.5 | 30.0 | 12.3 | 0.10 | - | - | - | 22.9 | 3.3 | 0.07 |
| | 77.1 | M | 177.8 | 88.9 | 24 | 35.2 | 33.4 | 221.8 | 0.14 | 28.6 | 7.9 | 0.19 | - | - | - |
| | 78.5 | M | 177.8 | 96.5 | 27 | 38.3 | 30.3 | 2.3 | 0.10 | 32.7 | 1.5 | 0.11 | - | - | - |
| | 78.5 | M | 180.3 | 95.3 | 36 | 38.1 | 40.3 | 170.4 | 0.16 | 28.8 | 77.6 | 0.10 | 27.8 | 13.0 | 0.09 |
| | 81.6 | M | 175.3 | 87.6 | 30 | 37.9 | 33.8 | 141.1 | 0.13 | 28.2 | 333.4 | 0.14 | 25.2 | 38.2 | 0.09 |
| | 81.6 | F | 157.5 | 87.0 | 23 | 36.8 | 38.4 | 18.4 | 0.13 | 52.6 | 202.1 | 0.20 | 29.2 | 13.8 | 0.10 |
| | 83.0 | F | 172.7 | 90.2 | 29 | 35.9 | 41.8 | 89.4 | 0.15 | 25.8 | 6.1 | 0.08 | 22.5 | 4.8 | 0.07 |
| | 83.0 | M | 185.4 | 97.2 | 28 | 38.2 | 35.8 | 430.6 | 0.18 | - | - | - | - | - | - |
| | 84.8 | M | 180.3 | 94.0 | 31 | 39.2 | 28.9 | 393.0 | 0.15 | 35.2 | 238.3 | 0.15 | - | - | - |
| | 88.5 | M | 182.9 | 96.5 | 27 | 38.8 | 31.5 | 129.5 | 0.12 | 39.6 | 4.5 | 0.13 | 23.3 | 11.4 | 0.08 |
| | 89.8 | M | 185.4 | 95.3 | 22 | 36.8 | - | - | - | - | - | - | 36.2 | 23.9 | 0.12 |
| | 90.7 | M | 181.6 | 96.5 | 37 | 39.7 | 37.0 | 429.8 | 0.18 | 34.0 | 3.2 | 0.11 | 26.6 | 88.3 | 0.10 |
| | 90.7 | M | 182.9 | 96.5 | 36 | 39.8 | 35.5 | 3.8 | 0.09 | 39.9 | 9.1 | 0.10 | - | - | - |
| | 99.8 | M | 189.2 | 99.1 | 33 | 39.8 | 48.8 | 228.8 | 0.15 | 28.5 | 449.4 | 0.12 | - | - | - |
| | 119.7 | M | 185.4 | 97.8 | 36 | 45.3 | 42.3 | 0.1 | 0.10 | 39.4 | 1.6 | 0.10 | - | - | - |
| | 126.1 | M | 193.0 | 97.8 | 32 | 45.0 | 64.6 | 10.1 | 0.16 | 41.8 | 72.4 | 0.11 | 28.1 | 7.1 | 0.07 |
| **Mean** | 80.4 | N/A | 175.6 | 92.3 | 29.1 | 36.6 | 37.7 | 162.6 | 0.14 | 34.2 | 148.0 | 0.14 | 26.1 | 39.3 | 0.09 |

Empty cells signify that the subject did not participate in designated experimental condition

*Procedures*

During the test, volunteer subjects were seated vertically and restrained in a standard AF fighter aircraft ACES-II ejection seat. Subjects were instructed to "brace." Bracing is a technique taught in pilot training to use the neck muscles to force the head back into the head rest, as it is believed this action reduces neck injuries due to forward flexion. The seat was mounted to the test sled and subjects were accelerated rearward on the sled track at the specified acceleration level to measure the $-G_X$ neck responses. A tri-axial linear accelerometer and an angular accelerometer mounted on a bite bar measured the head accelerations (Doczy et al.,

2004). The accelerative portion of the experiment lasted for about 200 ms. All of the tests were non-injurious but neck stiffness or soreness (classified as less than AIS 1 injuries) was reported in approximately 15% of the tests, mostly at the higher helmet mass and acceleration levels (Doczy et al., 2004). This post-test reporting was used by medical observers to determine subject safety and there were no clinical outcomes. All subjects had radiological scans taken before admittance to the subject panel and were cleared of any musculoskeletal and other pathological issues (e.g. observations of degenerative disks or osteoporosis) that would preclude them from participation in the study. Upon exit from the study panel, subjects typically underwent a brief survey performed by the medical examiner to check for pain or discomfort caused by the testing. If warranted, follow on radiological scans were performed. If not, the subjects were released. These actions could be after several test series and were not indicative of any particular test, but detailed the effects of many tests that possibly could result from years of exposures.

In this experimental paradigm, the expected kinematic response is for the head to flex forward at some point during the frontal impact and then transition into combined tension and flexion. Thus only peak $N_{TF}$ values were used when analyzing and making comparisons between each different experimental set up. Any other observed head and neck loading, like high compression values or other unexpected spikes in the $N_{ij}$ values near the end of the test were considered artifacts of the test attributed to the decelerating sled and thus not used in the analysis. In the lower acceleration test (6 Gs) some subjects were able to maintain a sufficient brace through the impact to prevent forward flexion. Neck load data for these subjects showed their necks never experienced the expected tension-flexion combination and thus their data was not applied in this analysis.

Independent variables for this research included helmet mass and acceleration, as well as individual anthropometric parameters of the subjects. The dependent variables were resultant head, neck and body accelerations which were used to compute the neck loads used in the $N_{ij}$ criteria (tension, compression, flexion, and extension). Neck and head mass was calculated using anthropometric measurements from each subject combined with separate regression equations from the literature for male and female neck volume and neck density values (Gallagher et al., 2007). Human subject neck loads were computed using subject anthropometry, exact helmet inertial properties, and bite bar recorded head accelerations at ms increments using a program used in previous studies (Doczy et al., 2004; Gallagher et al., 2007). This program is accurate for predicting forces during times of non-contact, thus the initial portion of the test when the subject is bracing is not accurate but these values were not used in this analysis since the peak loads occurred during peak acceleration of the head. At peak acceleration the head is off the headrest and not in contact with any other structures so it becomes purely an inertial calculation. The program does not consider the internal motion of brain tissue and other soft fluids, but assumes the head behaves as a rigid body. While it is understood these calculated force values from acceleration vectors are not exact, they are of adequate accuracy for the purposes of further understanding human neck response to acceleration. $N_{ij}$ values were subjected to statistical analysis to determine the sensitivity of $N_{ij}$ to changes in acceleration and helmet mass; as well as changes in anthropometric characteristics of the participants.

These same data were also applied to generate alternative AIS 2 and 3 human risk curves that are more appropriate for military aviation. In this portion of the analysis, the $N_{ij}$ data from these three human subject test conditions (n=67) were combined with a set of injurious PMHS $N_{ij}$ data (n=6) and risk curves were produced using a survival analysis. The six whole specimen PMHS data points were taken from previous research published by Cheng et al. (Cheng et al.,

1982). This data set provides the largest published, whole specimen, frontal impact research available which included both observed neck loads and injury level. Since this research was focused on injury risk curves generated from human and PMHS data, no data from matched paired PMHS/Hybrid-III tests were used. Frontal impact acceleration levels in this experiment were between 32 and 39 Gs. Peak observed neck loads were estimated using acceleration and head mass to calculate forces. Injury caused by the impact was determined by autopsy and specified on the AIS scale. Of the six PMHS, four experienced injuries classified as AIS 2 or greater, and three experienced injuries classified as AIS 3 or greater (Cheng et al., 1982). Thus the risk curve generated for AIS 2 injury and the risk curve generated for AIS 3 injury differ by a single injurious data point.

The $N_{ij}$ values used for the regression for the human subjects were the peak instantaneous value of the combined axial and bending loads. Unfortunately no time history was published for the PMHS data. Thus, only the peak individual values were reported and applied for axial and bending loads. Note that these forces did not necessarily occur at the same time. Because of this, the injurious $N_{ij}$ values are potentially higher than the peak instantaneous values specified by the NHTSA $N_{ij}$ construct. Thus, the resultant risk curve is slightly biased towards higher $N_{ij}$ values.

The $N_{ij}$ values were calculated using the published NHTSA $N_{ij}$ intercept values (Eppinger et al., 2000) based upon occupant size by applying the small sized female intercept for subjects with body mass less than 63.5 kg the mid-sized male intercept values for subjects with body mass between 63.5 kg and 90 kg and the large male intercept values for subjects with body mass greater than 90 kg.

*Statistical Analysis*

Risk curves were generated through parametric survival analysis (Hosmer et al., 2008) following the methods used in research by Bass et al. (Bass et al., 2006). Survival analysis has recently been proposed as the standard in the biomechanics field for generating injury risk curves over the traditional logistic regression approaches which were used to generate the original risk curves associated with $N_{ij}$ due to the ability of survival analysis to handle censored data (Cutcliffe et al., 2012). Using inverse prediction, the NHTSA and human data generated $N_{ij}$ risk curves were compared at the 5% and 22% risk levels. As noted earlier, the 5% risk level is significant to military aviation and the 22% risk level is significant to the NHTSA application of the $N_{ij}$ risk criteria.

**Results**

To assess the sensitivity of $N_{ij}$ to acceleration and helmet mass, the distributions of the $N_{ij}$ values for each test case were analyzed. Data from the three tests were moderately skewed, thus nonparametric statistical methods were applied. Additionally, since each test case used overlapping pools of subjects, the samples were not independent and thus the Related-Samples Wilcoxon Signed Rank Test was applied to compare $N_{ij}$ values across the three acceleration and HMD mass values as well as within each test case between various groups of individuals. A statistically significant difference in the $N_{ij}$ was observed when the acceleration was increased from 6 G to 8 G while the HMD mass of 2 kg was held constant (Related-Samples Wilcoxon Signed Rank Test $p=0.002$, $\alpha=0.05$, mean $N_{ij}$ of 0.0931 at 6 G and 0.138 at 8 G). When acceleration was held constant at 8 Gs and the HMD mass was varied from 1.6 kg to 2 kg, the

difference in $N_{ij}$ was not statistically significant (Related-Samples Wilcoxon Signed Rank Test $p=0.550$, $\alpha=0.05$, mean $N_{ij}$ of 0.136 at 1.6 kg and 0.138 at 2 kg).

Mean, as well as maximum and minimum $N_{ij}$ values for each condition are shown in **Figure 14**. $N_{ij}$ is lowest for the 6 G, 2 kg condition and increased as the acceleration was increased from 6 to 8 Gs. The effect of changing the helmet mass from 1.6 to 2 kg also affected the mean value slightly in the expected direction (e.g., mean $N_{ij}$ was slightly lower for the 1.6 kg helmet than the 2 kg helmet). However, at an acceleration of 8 Gs, the 0.4 kg change in helmet mass had a near negligible effect on mean $N_{ij}$.

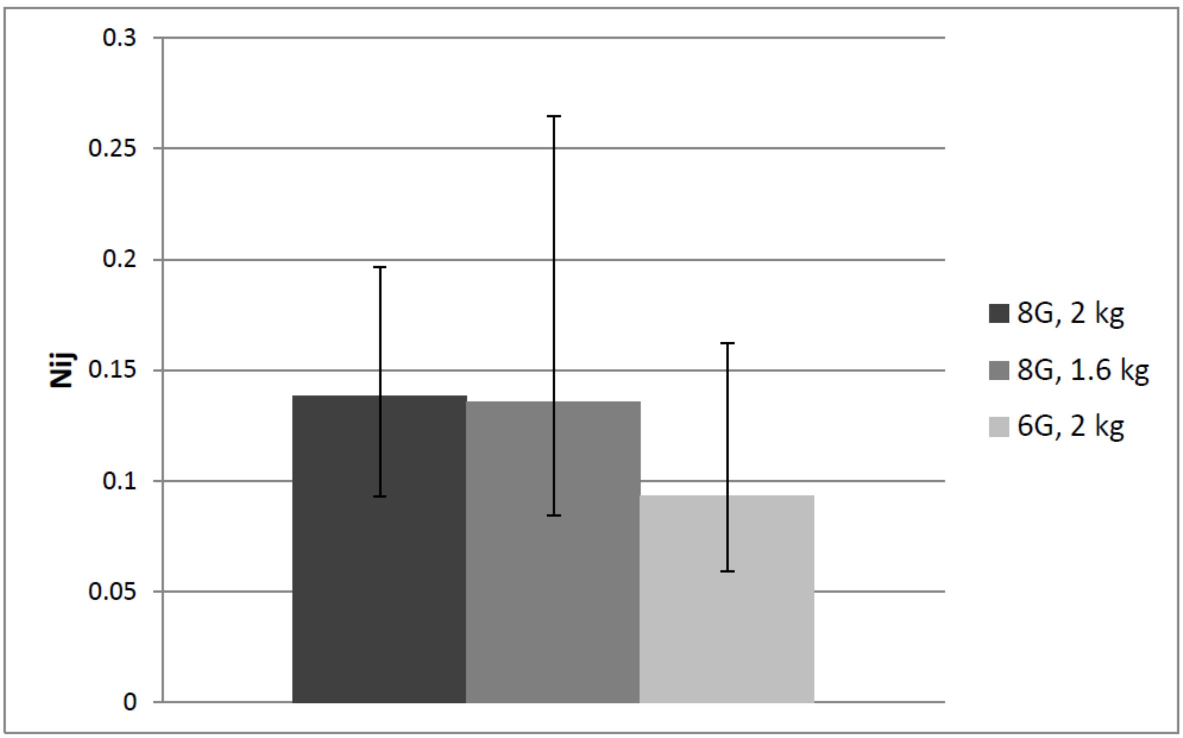


**Figure 14. Mean Nij Values Shown as a Function of Each of the Conditions (Error Bars Show Minimum and Maximum Values)**

Specific anthropometric factors were analyzed to determine if they contributed to the observed neck responses. Female peak instantaneous $N_{ij}$ values were not statistically different from male $N_{ij}$ values in any of the three conditions (p-values ranged from 0.31 to 0.89). The effect of body mass on human neck response was also investigated. The average body mass of all subjects was approximately 80 kg. The neck response for subjects whose mass was above the mean (80 kg) were compared with the subjects with less than average body mass. The independent samples Mann-Whitney U test indicated that no significant difference existed between the means for any of the three conditions (p values ranged from 0.14 to 0.96). The effects of sitting height and neck circumference on neck response were also investigated using a similar method of dividing the group based upon the mean. Neither measurement had a statistically significant difference on the mean for any of the three experimental conditions, with the exception of sitting height in the 8G/1.6kg condition where subjects with low sitting height experienced higher $N_{ij}$ values (p-values ranged from 0.016 to 0.85). Spearman's rank correlation was computed to determine the correlation between the $N_{ij}$ values and the anthropometric conditions of body mass, sitting height, and neck circumference for each test setup. For the 8G/2kg condition, correlation of $N_{ij}$ on body mass, sitting height, and neck circumference were -0.08, -0.25, and -.07 respectively; for the 6G/2kg condition, correlations were -0.10, -.25, and -0.19; and for the 8G/1.6kg condition correlations were -0.40, -0.55, and -0.34. No correlation between the anthropometric variables and $N_{ij}$ were statistically significant at a confidence level of 0.05 with the exception of the effect of sitting height in the 8G/1.6 kg condition.

Air Force aviation has required that a pilot have a 5% or less probability of an AIS 2 or greater injury during ejection. The relevant NHTSA risk curve is shown in **Figure 15**. Unfortunately, the NHTSA risk curve does not provide a 5% prediction as it intercepts the Y-axis at 11.3%. Therefore, to understand the $N_{ij}$ value that corresponds to the desired risk level, it

is necessary to generate an alternate risk curve. Towards this end, a revised risk curve was generated using survival analysis, combining data from 67 human subjects in a single frontal impact experiment with 6 PMHS from a separate, but similarly structured, frontal impact experiment to obtain the Human Risk Curve shown in **Figure 15**. As shown, the Human Risk Curve predicts a probability of injury at $N_{ij} = 0$ of only 0.52%, which is closer to the expected value of zero than the 11.3% probability produced by the NHTSA AIS 2 risk curve. Although the NHTSA risk curve predicts a 13% risk of AIS 2 or greater injury for the 8 G, 2 kg helmet condition mean $N_{ij}$ of 0.138, no AIS 2 injuries were observed in the human subject population. The AIS 2 or greater Human Risk Curve produced predicts a more accurate 0.91% risk of injury under these conditions. Additionally, the Human Risk Curve indicates that the probability of neck injury increases much more rapidly as a function of $N_{ij}$ than the NHTSA curve, reaching an asymptote near 100% probability at a $N_{ij}$ of 3 as opposed to 6 for the NHTSA curve. Also shown in **Figure 15** is the 95% confidence interval for the Human Risk Curve. Note that the NHTSA risk curve provides $N_{ij}$ values outside of this confidence interval for values below 0.51 and greater than 1.85. Using inverse prediction, a 5% risk of AIS 2 neck injury using the human data risk curve gives an $N_{ij}$ of 0.56 (95% confidence intervals of 0.129 and 0.998 respectively). The equation for the human AIS 2 risk curve is below.

$$P(\mathrm{AIS} \geq 2) = \frac{1}{1 + e^{5.2545 - 4.1 * N_{ij}}} \tag{4}$$

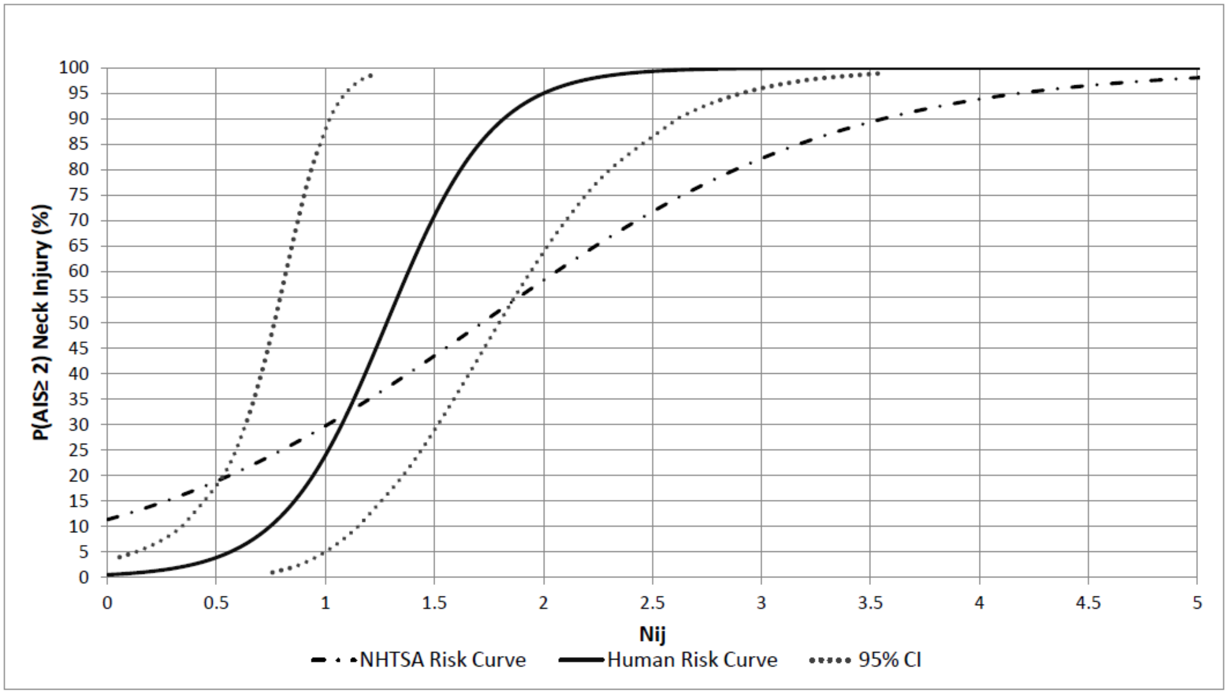


**Figure 15. Probability of AIS 2 or Greater NHTSA and Human Nij Neck Injury Risk Curves (95% CI Shown for Human Risk Curve)**

NHTSA applied the AIS 3 risk curve to determine the $N_{ij}$ performance limit for advanced automotive restraint systems, and thus it is beneficial to compare their AIS 3 risk curve to a human subject data generated risk curve at this same AIS 3 level (see **Figure 16**). The NHTSA AIS 3 risk curve predicts 3.8% risk of AIS 3 neck injury or greater at zero input, thus it is better at predicting lower levels of risk compared to the NHTSA AIS 2 risk curve. Once again, a revised risk curve was generated using survival analysis, combining 67 human subjects from a single frontal impact experiment with 6 PMHS to obtain the Human Risk Curve shown in **Figure 16**. Unlike the results obtained for the AIS 2 curve, most of the NHTSA AIS 3 risk curve lies within the 95% confidence interval generated for the revised Human risk curve, with the exception of $N_{ij}$ values below 0.2. Using inverse prediction, a 5% risk of AIS 3 neck injury using the human data risk curve gives an $N_{ij}$ of 0.72 (95% confidence intervals of 0.165 and

1.274 respectively). A 22% risk of AIS 3 injury using the human data risk curve gives a $N_{ij}$ of 1.23 (95% confidence intervals of 0.635 and 1.82 respectively) as compared to a $N_{ij}$ of 1.0 for the NHTSA risk curves. As such, it would appear that the human data risk curve provides an intercept nearer the expected value of 0 and provides a less conservative estimate of risk than the NHTSA risk curves for a specified 22% risk of AIS 3 injury or greater. The equation for the human AIS 3 risk curve is below.

$$P(AIS \geq 3) = \frac{1}{1 + e^{5.31423 - 3.3922 * N_{ij}}} \tag{5}$$

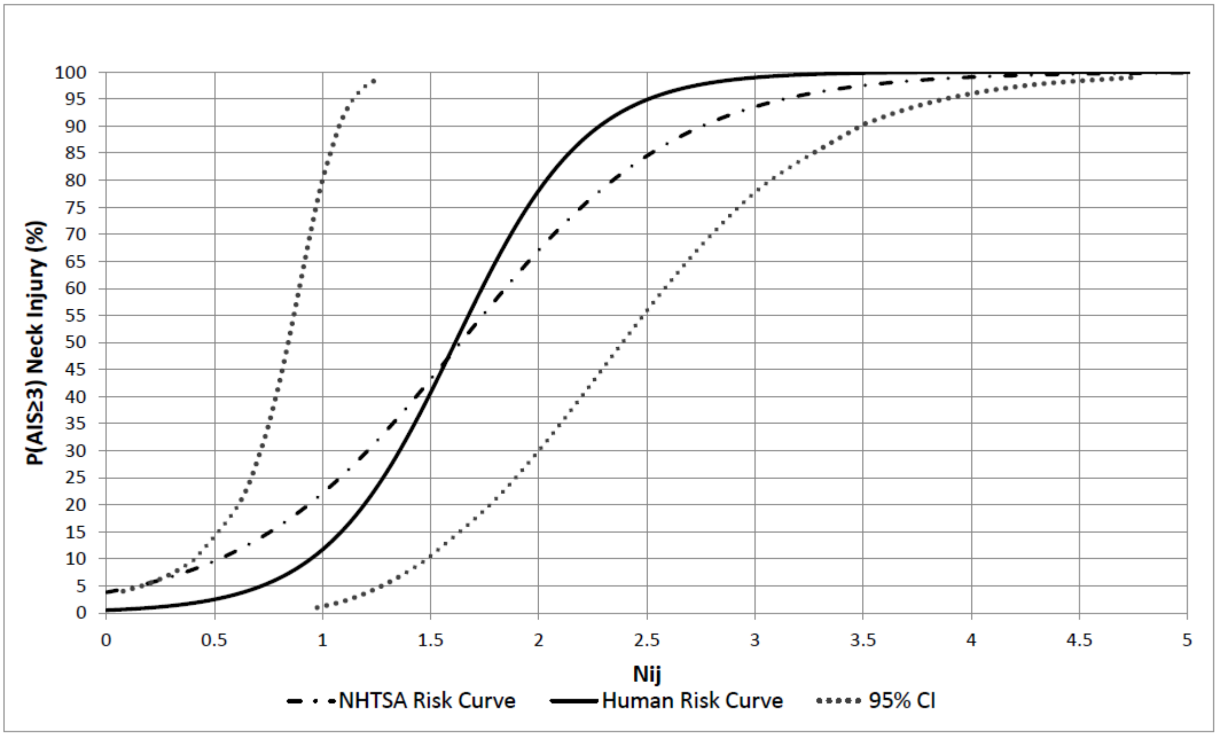


**Figure 16. Probability of AIS 3 or Greater NHTSA and Human Nij Neck Injury Risk Curves (95% CI Shown for Human Risk Curve)**

A comparison of the human data generated AIS 2 risk curve and the AIS 3 risk curve is provided in **Figure 17**. As stated previously the difference observed in the AIS 2 and 3 risk curves is produced by a single injury data point in the source data, indicating the sensitivity of the injury criteria when the sample size for the PMHS is small, as in this data set. These curves behave as would be expected. At the higher injury level, a greater value for $N_{ij}$ is allowed at a specific risk level. For example, at 5% risk of injury, the AIS 2 risk curve allows for an $N_{ij}$=0.56 and the AIS 3 risk curve allows for an $N_{ij}$=0.72.

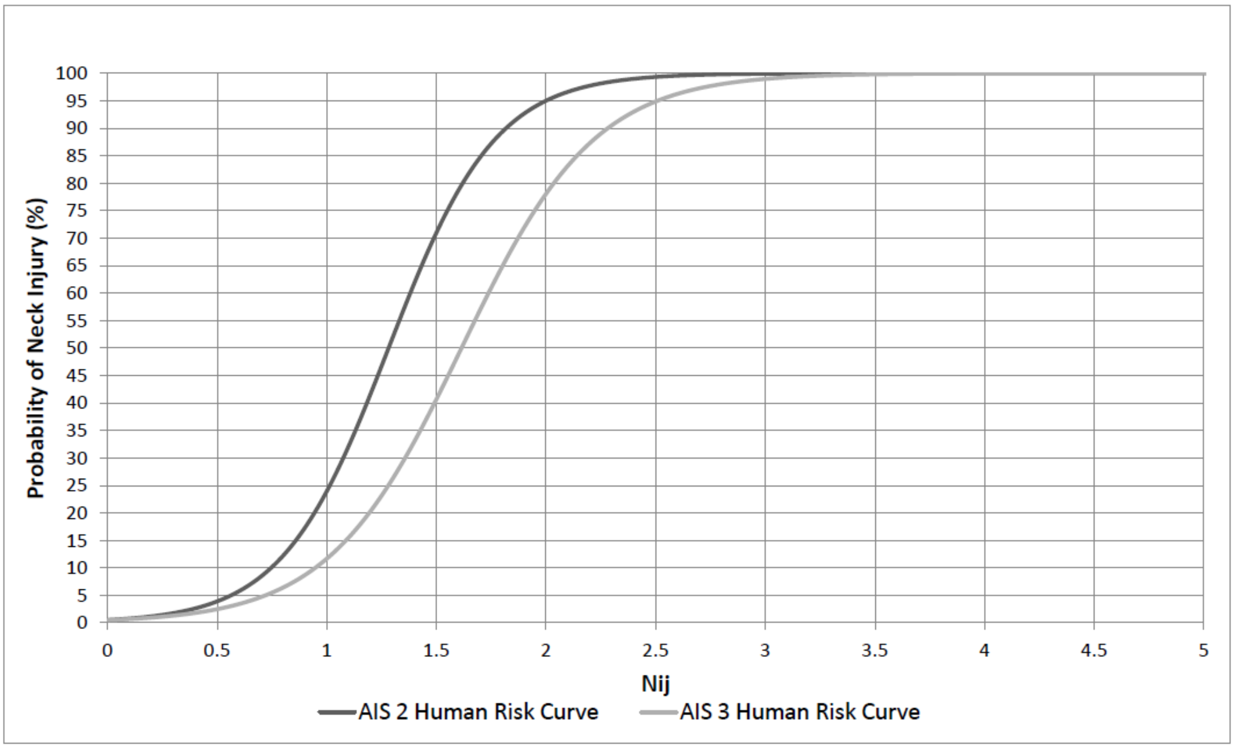


**Figure 17. Probability of AIS 2 or Greater and AIS 3 or Greater Human Nij Neck Injury Risk Curves**

## Discussion

This study sought to assess the applicability of the $N_{ij}$ criteria to the evaluation of helmet systems of varying mass under various acceleration levels as well as to compare the NHTSA $N_{ij}$ risk curves to human data constructed $N_{ij}$ risk curves. When considering the neck response forces used in the $N_{ij}$, this study found that an increase in acceleration by 2 Gs had a greater impact on neck response than an increase in HMD mass of 0.45 kg. Although the change in helmet mass did not have a significant effect on $N_{ij}$, it is not clear whether this result is appropriate since the mass difference of the two test HMDs was relatively small. Whether this change in mass has a negligible effect on injury risk at the given acceleration levels or whether the $N_{ij}$ does not appropriately account for an increase in risk requires further investigation.

Based upon the construct of the $N_{ij}$, which includes critical intercept values that normalize the criteria based upon varying occupant size, $N_{ij}$ would not be expected to vary significantly based upon anthropometric differences related to size. That is, if the NHTSA intercept values are accurate, there should not be a statistically significant difference in $N_{ij}$ due to gender, body mass, neck circumference, or sitting height. This study showed that the NHTSA intercept values did an adequate job of normalizing neck response for subject anthropometry based upon the observation that body mass, sitting height, and neck circumference were not significantly correlated with $N_{ij}$ in any of the three test configurations, with the exception of sitting height in the 8 G, 1.6 kg test. Further supporting the finding that the critical intercept values satisfactorily compensate for anthropometric factors was the finding that subject gender did not have a significant effect on the resultant neck loads. Further, the neck loads were found not to be statistically significant between individuals with greater or less than average mass, neck circumference, or sitting height, with the exception of sitting height in the 8 G, 1.6 kg test.

The NHTSA $N_{ij}$ neck injury criteria used paired piglet/ATD data to determine neck load and assess injury/no injury and then scaled this criteria to estimate human injury. However, other approaches have been applied. For instance, the AF tensile neck injury criterion combined data from non-injurious human subject data with PMHS injury data to construct a risk curve (Carter et al., 2000). In formulating these curves, human data (n=208) was used for the non-injurious neck load data points and the PMHS data (n=10) was used for the injurious neck load data points. The FAA has applied other methods, including pairing injury data from PMHS (n=10) with the neck load data from an ES-2 ATD to create tensile neck injury criteria for qualifying side facing aircraft seats (FAA, 2011). Each of these approaches has advantages and disadvantages. The combined human and PMHS method used in the AF tensile neck injury criterion has the advantage of a greater sample size (n=218 vs n=10) compared to the PMHS only used by the FAA, which provides greater statistical power. It also directly estimates the neck load the subject experienced rather than assuming that the paired piglet/PMHS and ATD tests resulted in equivalent neck loading scenarios as the NHTSA $N_{ij}$ and FAA side-facing seat methods assume. The disadvantage of the paired piglet/ATD and PMHS/ATD methods are the relatively small samples sizes in studies using PMHS based upon the availability and suitability of subjects. This small sample size makes statistical significance of the risk function and resultant injury criteria an issue for use as a predictive tool.

As NHTSA's risk curves are not useful to determine the $N_{ij}$ value for a 5% risk of AIS 2 injury as required for military aviation, it was necessary to generate revised Human Risk Curves. A summary of the predicted $N_{ij}$ values and $N_{ij}$ values from the Human AIS curves at key risk values are provided in Table 8. The fact that the NHTSA curves were constructed with a smaller number of low $N_{ij}$ values resulting in no injury appears to have resulted in AIS 2 and AIS 3 curves which over predict risk of injury at lower $N_{ij}$ levels. Conversely the human risk curves

were constructed with many data points at lower $N_{ij}$ values, which resulted in AIS curves which indicate human tolerance at moderate $N_{ij}$ levels. Applying the AFLCMC escape system oversight office recommended 5% limit to the AIS 2 human risk curve would result in a maximum allowable $N_{ij}$ value of 0.56. Although this value is relatively close to the performance limit of 0.5, which is currently being applied in this domain (Nichols, 2006) with existing ATDs, as the limit calculated here has not been cross correlated with ATD response, caution should be taken when comparing these numbers.

**Table 8. Risk Curve Prediction Values**

| Risk Curve | $N_{ij}$=0 Injury Prediction | 5% $N_{ij}$ Prediction | 22% $N_{ij}$ Prediction |
|---|---|---|---|
| NHTSA AIS 2 | 11.3% | N/A | 0.66 |
| Human AIS 2 | 0.52% | 0.56 | 0.97 |
| NHTSA AIS 3 | 3.8% | 0.114 | 1.0 |
| Human AIS 3 | 0.49% | 0.72 | 1.23 |

This research analyzed different methods of constructing risk curves. For the combined human/PMHS method it was highlighted that for more statistical significance of the risk curves and resultant injury criteria, more PMHS testing is needed, with time history neck load data and injuries specified at the specific AIS levels. It is recommended that the test setup for the human and PMHS experiments be as close as possible, varying only input acceleration levels to achieve injurious results with the PMHS. Ultimately, this course of research might lead to an aviation specific, human data supported, neck injury criteria that would not only evaluate prototype HMD designs but also provide design guidance parameters for the mass properties of future HMDs.

A few limitations of this study are worth noting. One issue in the area of human subject testing in accelerative environments is the use of small sample sizes. Testing of this kind is expensive, requires very comprehensive medical screening of volunteer subjects, and in some cases subjects remove themselves voluntarily from further testing for a variety of reasons,

including neck discomfort. For example, of the 34 human subjects (16 females and 18 males) that participated in this particular -Gx accelerative study, results were gathered for 9 females and 15 males for the 8 G, 1.6 kg HMD test, 9 females and 17 males for the 6 G, 2 kg test, and 7 females and 16 males for the 8 G, 2 kg condition. The power of the intra-sample comparisons would have been greater if all subjects participated in all conditions. The overall sample size for the three test runs of 23, 24, and 26 subjects was further reduced when the group was divided to permit comparison of the effects of gender, body mass, neck circumference, and sitting height, further reducing the power of the statistical tests. In addition, the small number of PMHS injurious data points involved in the regression results in a statistically underpowered curve to be used to predict risk of neck injury. This study should be seen as a pilot study and additional injurious data should be included in the generation of the injury risk function before attempting to apply the curve to real world risk predictions. Additionally, this study only used human neck response data to generate the injury risk functions and did not attempt to relate neck loads observed in the Hybrid-III or other ATD with human neck injury as is done in the traditional application of the $N_{ij}$ criteria. Since the Hybrid-III neck has been observed to be non-biofidelic and not sensitive when used with head mounted mass (Bass et al., 2006; Salzar et al., 2009), application of the revised injury risk curves developed in this paper with a better suited ATD is necessary to apply this research in system evaluation. Furthermore, based upon the construct of the $N_{ij}$, this study considered only upper neck loads. Bass et al. found that added head supported mass resulted in different head and neck kinematics compared with an unloaded head, resulting in greater injury potential to the lower neck (Bass et al., 2006). Future aviation-specific neck injury criteria should consider and potentially incorporate loading of the lower neck.

This paper advances knowledge in this area of study in two ways. First, by applying the $N_{ij}$ to human subject data, important observations were made as to the sensitivity and

appropriateness of this neck injury criterion to helmet mass, acceleration, and anthropometric factors.  Second, generating injury risk curves using combined human and PMHS neck load data allowed for fruitful comparison and evaluation of the appropriateness of the NHTSA injury risk curves in the ejection environment.

The $N_{ij}$ construct shows potential for use as an evaluative tool for HMD and escape system development as it embodies key characteristics, including a method to account for anthropometric differences and the ability to link probability of injury with restraint and helmet system imposed differences in neck response for at least conditions similar to frontal automotive crashes or the parachute shock portion of ejection.  As a result, a revised form of this criterion evaluated through a more biofidelic ATD neck than the Hybrid-III may be useful as a tool to evaluate the overall neck load impact of different HMD loading conditions and different accelerations applied in the evaluation of new HMDs.  Unfortunately, the $N_{ij}$ is reactive rather than proactive when guiding HMD mass properties.  That is, the criteria will provide information related to the acceptability of a fully prototyped HMD or escape system, but in its current format does not provide guidance to inform the design process.  Besides the need to better understand the impact of helmet mass on this criterion, further advances, including adjustment to the formulation to account for the forces that are likely to occur for the remaining three phases of ejection and the ability to extend this criterion to provide predictive engineering tools are fruitful areas for further investigation.  A larger scale study is now needed to further clarify these issues.

# V. Development of a Side Impact (Gy) Neck Injury Criterion for use in Ejection System Safety Evaluation

## Chapter Overview

The paper that comprises this chapter will be submitted for publication to the IIE Transactions on Occupational Ergonomics and Human Factors Journal. This paper outlines the development of a multi-axial side impact (Gy, see **Figure 18**) neck injury criterion using combined human subject and PMHS data. The Gy axis of acceleration is unique compared to Gx and Gz in that it is assumed to be equivalent from either direction; there is no differentiation between -Gy and +Gy as there is with Gx and Gz.

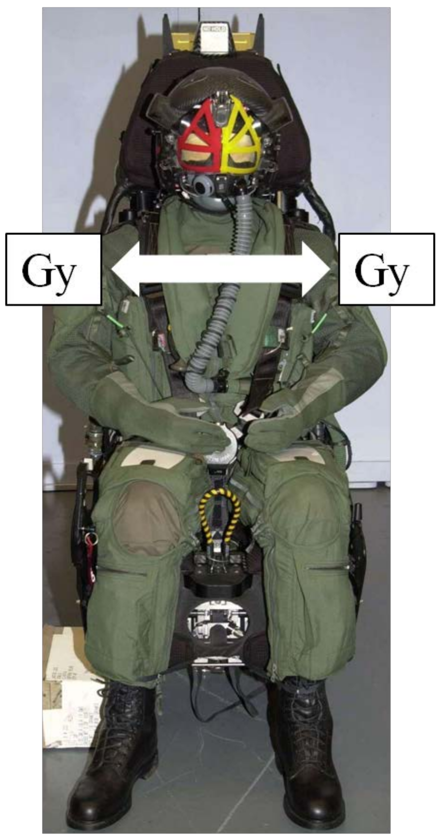


**Figure 18. Gy Axis of Acceleration**

## ABSTRACT

**BACKGROUND:** Ejection neck safety remains a concern in military aviation with the growing use of helmet mounted displays (HMDs) worn for entire mission durations. A lateral (Gy) impact upper neck injury criterion is developed and proposed to evaluate DoD escape systems and HMDs. These same criteria may be useful analyzing side impact safety in other vehicle systems. **METHODS:** A multi-axial lateral impact risk function (referred to as MANIC(Gy)) was constructed with a combined human subject (N=56) and post mortem human subject (N=9) data set using Survival Analysis. The human subject data were analyzed to observe criteria Gy sensitivity to anthropometric factors. The risk function was applied to quantify the risk associated with changes in HMD mass and acceleration input. **RESULTS:** An AIS 2 or greater, lateral impact (Gy), upper-neck, injury criterion is proposed, which yields a 5% risk of AIS 2 or greater injury at a criteria value of 0.48 (95% confidence intervals of 0.28 and 0.67 respectively). An AIS 3 or greater risk function was also generated, yielding a 5% risk of AIS 3 or greater injury at a criteria value of 0.53 (95% confidence intervals of 0.24 and 0.82 respectively). **DISCUSSION:** This pilot scale multi-axial risk function could be applied to quantify the risk of neck injury posed by lateral acceleration. Criteria values were correlated with body mass and other body mass related anthropometric factors, indicating that the critical values incorporated in this study may be inaccurate.

## Introduction

Injury risk posed by accelerative forces must be understood to evaluate the safety of powered vehicles. It is particularly important to understand this risk when developing a new aircraft or a re-designed escape system for legacy aircraft due to the high accelerative forces necessary for safe ejection under a broad range of air speeds. In modern aircraft, understanding risk is complicated both by the presence of head-born weight and increasing pilot anthropometric diversity.

Helmet mounted displays (HMDs) are becoming common human-machine interface equipment in manned flight. They have been shown to increase the performance of operators in their weapon systems and thus increase overall mission effectiveness by adding capabilities such as enhanced night vision and information fusion, which have the potential to enhance mission effectiveness across the spectrum of military operations (Rash et al., 2009). Unfortunately, this increased capability is often accompanied by increased mass, which can threaten pilot safety during ejection (Lewis, 2006; Nakamura, 2007; Stemper et al., 2009). Heavier HMDs worn for mission durations pose greater threat to the neck of pilots in an ejection than the lighter standard flight helmet. Of particular interest, the increased mass has the potential to increase the risk of operator neck injury if the pilot is subjected to accelerative environments like ejection. Injury risk due to a heavier HMD in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures, ligament ruptures, and spinal cord damage (Buhrman and Perry, 1994; Stemper et al., 2009).

Pilot anthropometric factors also affect the likelihood of injury from neck loads induced by acceleration and head supported mass. Recent changes in Department of Defense pilot accommodation requirements have increased the range of pilot size (Harris, 1997). Therefore, it

is important that pilot neck response to accelerative forces be understood and characterized using a standard evaluation criteria that considers the influence of pilot anthropometric characteristics.

To adequately protect pilot's necks to an acceptable risk of injury it is important to understand risk and develop risk functions for all axes of acceleration. The National Highway Transportation and Safety Administration employs a neck injury criterion called the Nij specially designed to limit injury in frontal impact (-Gx) automobile accidents. Previous studies have developed aviation specific risk functions and criteria for Gx acceleration which account for head supported mass (Bass et al., 2006; Parr et al., 2013). Gy, or side acceleration, can happen in ejection, particularly if the ejection seat turns with respect to the axis of aircraft travel which is possible in the highly dynamic ejection environment. Similar accelerations can be present in side impact collisions within other motorized vehicles. For military aviation, and many other domains, it is important to develop risk functions for all primary axes of acceleration from which to establish multi-axial neck injury criteria to aid the design and testing of new escape and HMD systems. This paper analyzes human subject upper neck (occipital condyles or OC) data from Gy acceleration experiments with head supported mass using a multi-axial neck injury criteria formulation. It also describes a method to develop a risk function for the Gy axis of acceleration using combined human and post mortem human subject (PMHS) data. It is proposed that this risk function might serve as the basis for a side impact risk criterion due to the fact that it was designed to meet Air Force escape system injury criteria requirements for application to accelerative environments. This work aims to address the following research questions:

1) What is a proper multi-axial neck load formulation?
2) Is a multi-axial neck load formulation applying previously determined critical values sensitive to various anthropometric factors (gender, body mass, head circumference,

sitting height, height, and age) in human subject lateral accelerative loading with head supported mass?

3) Does a multi-axial risk function constructed with human subject and PMHS data demonstrate sensitivity to varying head supported mass and input acceleration?

Various methods of modern risk curve construction have been outlined in other work. Matched pair ATD and PMHS experiments were used to construct the FAA's neck injury criteria for side-facing aircraft seats (FAA, 2011). The AF tensile neck injury criterion and a proposed modified Nij for use in the aviation ejection environment both incorporate the method of combining human subject and PMHS data (Carter et al., 2000; Parr et al., 2013; Pellettiere, 2012). NHTSA's widely used Nij upper neck injury criteria used matched pair ATD and piglet experiments scaled to human applicability to evaluate automotive restraint systems in frontal crash scenarios (-Gx accelerative input) (Eppinger et al., 1999). The Beam Criterion, a modification of the Nij for the lower neck for use with head supported mass, used data collected from instrumented PMHS neck sections (Bass et al., 2006). The current research employs the method of combining human subject and PMHS data. The main benefits of this method are that accurate human neck loading and injury are observed and that these observations are incorporated directly into the risk function. These two benefits are arguably the most important elements of an accurate and applicable risk function.

There are also some limitations to the combined human subject/PMHS method of risk curve development. In the field of injury biomechanics, experiments with both human subjects and PMHS are often expensive and data are difficult to collect, resulting in relatively small data sets. Neck response to non-injurious loading can be collected from human subject testing, but the loading is estimated from observed head acceleration data and head/helmet inertial properties combined with subject anthropometry (Parr et al., 2013). Neck response and injury data

collected from experiments with PMHS may not be representative of the typically young, fit military flying population and lacks active musculature, potentially resulting in overly conservative criteria. Human subject testing requires extensive approval procedures from Institutional Review Boards, and PMHS testing is limited to available specimens that require careful storage, handling, and injury assessment procedures (e.g. necropsy by trained personnel and radiographic scans).

**Methods**

This study has two parts. First, Gy human subject neck response data from variable helmet mass and accelerative input was analyzed statistically to see if differences were observed in neck loads based upon gender, body mass, head circumference, sitting height, height, and age. Second, a risk function was constructed using a combination of this Gy human subject data paired with Gy PMHS data.

Two human subject data sets from a previous Gy acceleration experiment were used in this study. They were chosen because they provided the highest lateral neck load exposure of the experiments that have been performed to date at the Air Force Research Laboratory (AFRL) horizontal impulse accelerator test facility. Subjects were restrained in an ejection seat representative of operational AF aircraft and subjected to a lateral (Gy), half-sine accelerative pulse with rise time and pulse duration of 75 and 150 ms respectively. The first study subjected 31 participants (21 male, 10 female) to 6 Gs of lateral acceleration (~5.5 m/s) with 1.36kg (3lb) of head supported mass. The second study subjected 25 subjects (17 male, 8 female) to 5 Gs of lateral acceleration (~4.6 m/s) with 2kg (4.5lb) of head supported mass. The typical kinematic response of the human subjects to Gy acceleration observed in slow motion video footage was an

initial combination of neck twisting moment (*Mz*) and coronal moment (side bending or *Mx*) with the addition of flexion (+*My*) to this combination near the end of the accelerative pulse. Pure coronal moment was not observed as a result of Gy acceleration.

Additionally, a PMHS data set was used in this study. These data are from research that supported the development of a neck injury criterion for side facing aircraft seats by a team of researchers from Medical College of Wisconsin, Wayne State, and the Federal Aviation Administration (FAA) (FAA, 2011). From this data set, time history upper neck load data was available from 9 PMHS experiments subjected to Gy acceleration that ranged from 8.5-19 G. The subjects were placed into one of three different test seating configurations representative of typical side-facing aircraft seats and restraints (FAA, 2011). Upper neck loads were calculated based upon observed head acceleration and subject anthropometry. Injury assessment post-test was categorized using the Abbreviated Injury Scale (AIS), and ranged from AIS levels 0 to 5 (AAM, 2008). For additional detail on the test set up, screening procedures and PMHS anthropometry the reader is referred to the final FAA summary report (FAA, 2011).

At the start of the study it was desired that a complete 6-load multi-axial structure be used for the independent variable of the risk function, which would include all six primary neck loads in a root sum of squares formulation as shown in Equation 6, called the multi-axial neck injury criteria (MANIC) after Perry et al. (Perry et al., 1997). The denominators for each component force that comprise the MANIC are critical values that scale each force based upon known component neck strength and occupant size. For example, the neck is stronger in flexion (+*My*) than extension (-*My*) and thus the critical value is higher for flexion than it is for extension (Eppinger et al., 1999). The use of critical values to achieve this scaling has been applied in other multi-axial structures, including the Nij (Eppinger et al., 2000). In a combined loading neck injury criterion structure such as the MANIC, the critical values are very important to

ensure that the proper weight is assigned and the individual contribution of each force relative to the other forces is captured in the criterion.

$$MANIC = \sqrt{\left(\frac{Fx}{Fxcrit}\right)^2 + \left(\frac{Fy}{Fycrit}\right)^2 + \left(\frac{Fz}{Fzcrit}\right)^2 + \left(\frac{Mx}{Mxcrit}\right)^2 + \left(\frac{My}{Mycrit}\right)^2 + \left(\frac{Mz}{Mzcrit}\right)^2} \quad (6)$$

where

*Fx* = observed x direction shear loading

*Fxcrit* = critical intercept value for x direction shear loading

*Fy* = observed y direction shear loading

*Fycrit* = critical intercept value for y direction shear loading

*Fz* = observed axial loading (+*Fz* = tension, -*Fz* = compression)

*Fzcrit* = critical intercept value for axial loading (different for tension/compression)

*Mx* = observed moment about the anatomical x axis (side bending)

*Mxcrit* = critical intercept value for side bending

*My* = observed moment about the anatomical y axis (sagittal plane anterior/posterior bending, +*My* = flexion, -*My* = extension)

*Mycrit* = critical intercept value for sagittal plane moments (different for flexion/extension)

*Mz* = observed moment about the anatomical z axis (neck twisting)

*Mzcrit* = critical intercept value for neck twisting

However, data availability necessitated a modified structure. The human subject experiment from which the data was collected was performed before bite bar sensors were small enough to accommodate accelerometers to observe all six primary OC neck loads (*Fx, Fy, Fz,*

*Mx, My,* and *Mz*). Due to the lack of observed *Mx* in human subject kinematics at the time of the original human subject experiment it was decided that angular acceleration about the x-axis would not be recorded. Thus *Mx* (side bending) data was not observed. As a result, a modified formulation with five of the six primary neck loads (*Mx* excluded) was used to compute the peak instantaneous MANIC as seen in Equation 7, referred to as MANIC(Gy).

$$MANIC(Gy) = \sqrt{\left(\frac{Fx}{Fxcrit}\right)^2 + \left(\frac{Fy}{Fycrit}\right)^2 + \left(\frac{Fz}{Fzcrit}\right)^2 + \left(\frac{My}{Mycrit}\right)^2 + \left(\frac{Mz}{Mzcrit}\right)^2} \qquad (7)$$

The critical values used in this study are values that have been used in a recent Department of Defense escape system qualification testing program (Nichols, 2006). These values incorporate data from the NHTSA Nij neck injury criteria (Eppinger et al., 2000) as well as Navy escape system qualification testing neck injury criteria (Nichols, 2006) and are scaled for ATD mass. Applying the ATD critical values to human subjects was done as described in **Table 9** (for example, 150 lb intercept values would be used for a human subject with a mass from 143 lbs to 161 lbs). The Nij has established critical values for +/- *Fz* and +/- *My*. For the forces that are not included in the Nij (*Fx, Fy, Mx, and Mz*), the critical values are based upon appropriate thresholds determined to limit injury in the ejection environment (Nichols, 2006). **Table 9** shows the intercepts used to calculate the MANIC(Gy) based upon subject body mass.

**Table 9. Upper Neck Critical Values Based Upon Body Mass**

| ATD Mass (lbs) | Human Mass (lbs) | Component | Force lb | Force N | Component | Moment in-lb | Moment N-m |
|---|---|---|---|---|---|---|---|
| 103 | <114 | $F_{xcrit}/F_{ycrit}$ | 405 | 1802 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 593 | 67 |
| | | $-F_{zcrit}$ (Comp) | 872 | 3880 | $+M_{ycrit}$ (flexion) | 1372 | 155 |
| | | $+F_{zcrit}$ (Tens) | 964 | 4287 | | | |
| 125 | 114-130.5 | $F_{xcrit}/F_{ycrit}$ | 496 | 2206 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 845 | 95 |
| | | $-F_{zcrit}$ (Comp) | 1099 | 4889 | $+M_{ycrit}$ (flexion) | 1939 | 219 |
| | | $+F_{zcrit}$ (Tens) | 1214 | 5400 | | | |
| 136 | 130.5-143 | $F_{xcrit}/F_{ycrit}$ | 522 | 2322 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 912 | 103 |
| | | $-F_{zcrit}$ (Comp) | 1157 | 5147 | $+M_{ycrit}$ (flexion) | 2094 | 237 |
| | | $+F_{zcrit}$ (Tens) | 1278 | 5685 | | | 0 |
| 150 | 143-161 | $F_{xcrit}/F_{ycrit}$ | 561 | 2495 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1016 | 115 |
| | | $-F_{zcrit}$ (Comp) | 1243 | 5529 | $+M_{ycrit}$ (flexion) | 2333 | 264 |
| | | $+F_{zcrit}$ (Tens) | 1373 | 6107 | | | 0 |
| 172 | 161-186 | $F_{xcrit}/F_{ycrit}$ | 625 | 2780 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1195 | 135 |
| | | $-F_{zcrit}$ (Comp) | 1385 | 6160 | $+M_{ycrit}$ (flexion) | 2744 | 310 |
| | | $+F_{zcrit}$ (Tens) | 1530 | 6806 | | | |
| 200 | 186-210 | $F_{xcrit}/F_{ycrit}$ | 683 | 3038 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1364 | 154 |
| | | $-F_{zcrit}$ (Comp) | 1513 | 6730 | $+M_{ycrit}$ (flexion) | 3133 | 354 |
| | | $+F_{zcrit}$ (Tens) | 1671 | 7433 | | | |
| 220 | 210-232.5 | $F_{xcrit}/F_{ycrit}$ | 777 | 3456 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1584 | 179 |
| | | $-F_{zcrit}$ (Comp) | 1673 | 7440 | $+M_{ycrit}$ (flexion) | 3673 | 415 |
| | | $+F_{zcrit}$ (Tens) | 1847 | 8216 | | | |
| 245 | 232.5+ | $F_{xcrit}/F_{ycrit}$ | 836 | 3719 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1850 | 209 |
| | | $-F_{zcrit}$ (Comp) | 1853 | 8243 | $+M_{ycrit}$ (flexion) | 4248 | 480 |
| | | $+F_{zcrit}$ (Tens) | 2047 | 9106 | | | |

### Statistical Analysis

First, a statistical analysis was performed to analyze the human subject anthropometric and MANIC(Gy) data. Non-parametric methods were used to compare male versus female neck response (in the SPSS statistics software package) and linear regression was used to assess the correlation of MANIC(Gy) with body mass, head circumference, sitting height, height, and age (in the JMP version 11 statistics software package). Finally, risk functions were constructed to predict AIS 2 or greater and AIS 3 or greater injury using survival analysis (SA). The time history data of each subject's accelerative test was processed. The Human participant data provided the sub-injurious data points for the risk function, while the FAA side impact PMHS data provided the injurious data points.

To determine MANIC values, the unitless MANIC(Gy) was computed at each step in the time history of each subject's test run. **Figure 19** shows an example non-injurious human subject MANIC(Gy) time history plot with a peak value of 0.22 which occurs at 130 ms. **Figure 20** shows an example injurious (AIS 5) PMHS MANIC(Gy) time history plot with a peak value of 1.6 which occurred at 97.2 ms. The head of the 160 lb subject struck the head rest at 102 ms, thus the plot is truncated at 102 ms. Any neck load values recorded after a head strike would be inaccurate due to the effect head impact has on the measured head accelerations used to calculate the neck loads. The appropriate intercepts based upon the subjects' body mass from **Table 9** were applied in the computation of the MANIC(Gy). Then, the peak MANIC(Gy) value and the corresponding level of injury observed during the test according to the AIS scale were used to generate a data set consisting of peak MANIC(Gy) values and injury assessment. In this case, an injury risk function built to evaluate AIS 2 or greater injury was desired. Thus the injury assessment was binary; either the subject did or did not experience an AIS 2 or greater injury. An AIS 3 or greater risk function was also constructed for comparative analysis purposes. An

AIS 3 or greater risk function might also be relevant to other fields (e.g. automotive) where a higher risk of injury may be tolerable due to the availability of emergency services. The survival analysis was performed on the combined human subject and PMHS data set to construct the risk function. SA accounts for censoring in the data (human subject data is right censored and PMHS data is left censored) and has recently been applied to the creation of human risk functions (Hosmer et al., 2008; Cutcliffe et al., 2012; Parr et al., 2013).

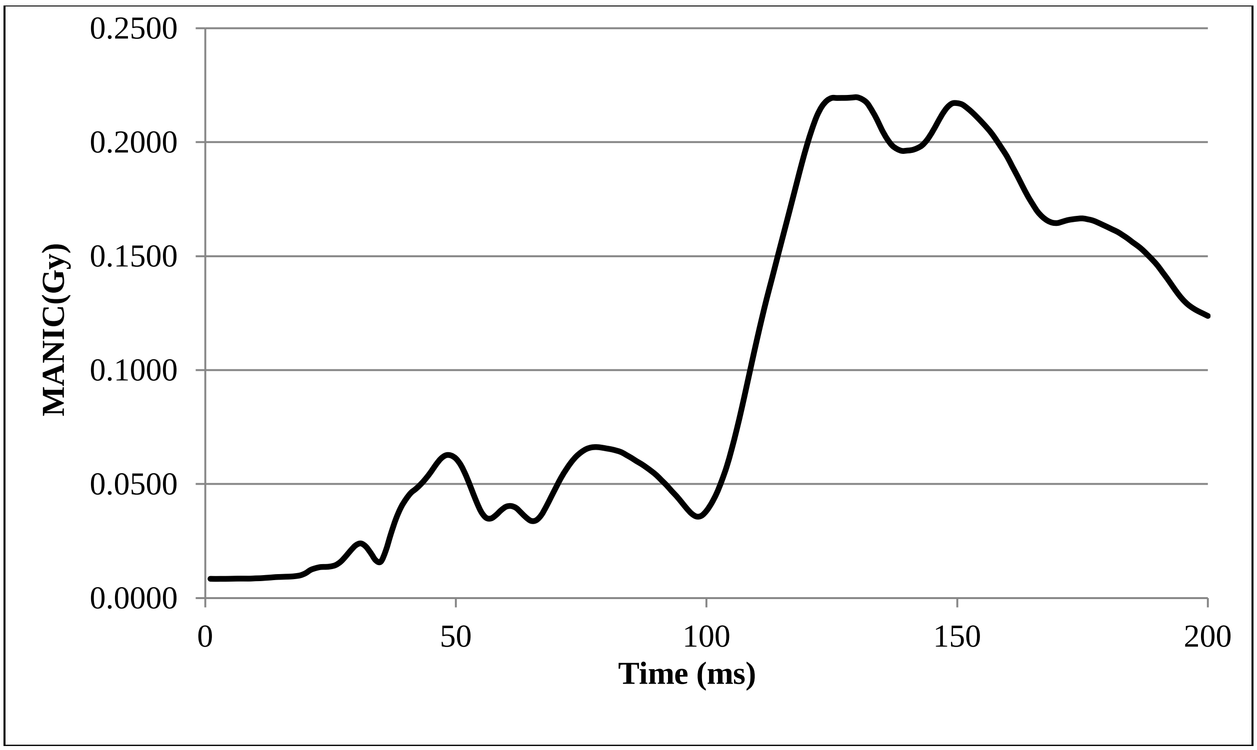


**Figure 19. Sample Human Subject MANIC(Gy) Time History Plot**

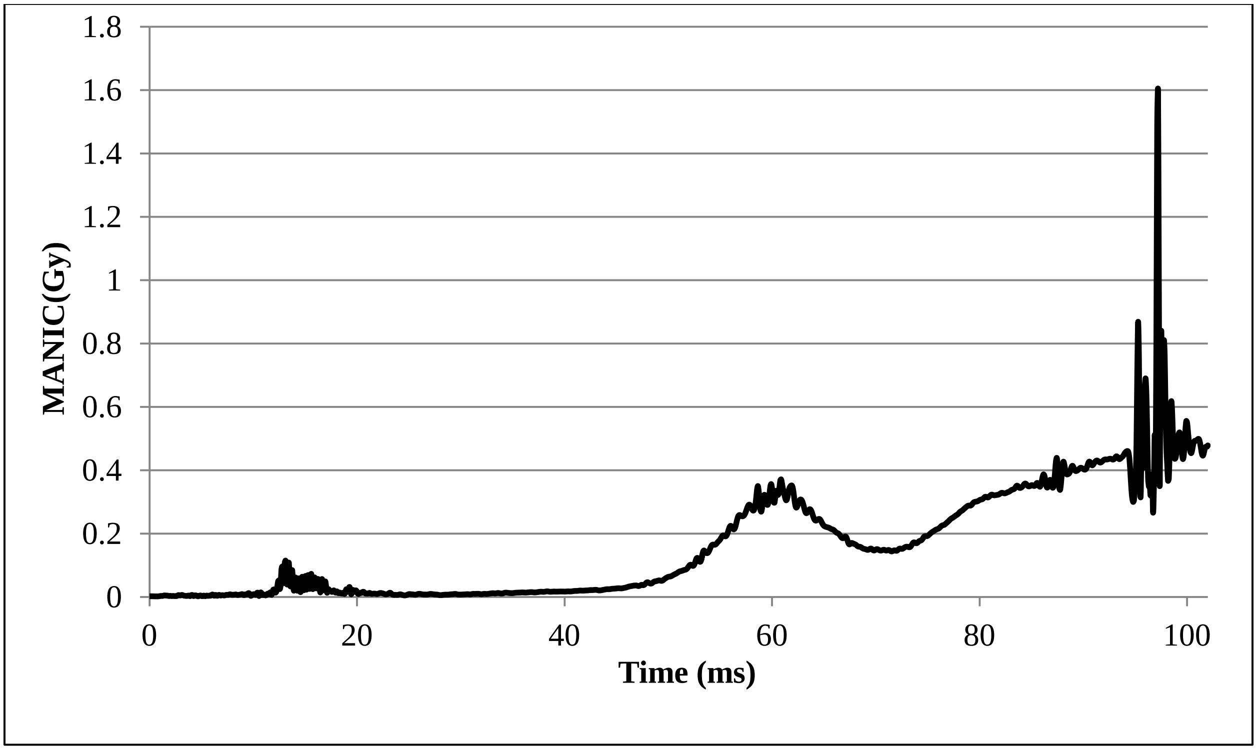


**Figure 20. Sample PMHS MANIC(Gy) Time History Plot**

**Results**

The human subject response data was analyzed and intra-test statistical analyses were performed. **Table 10** and **Table 11** show subject anthropometry and resultant MANIC(Gy) values for the 6G / 3lb (1.36kg) experimental conditions and the 5G / 4.5lb (2kg) experimental conditions respectively. Mean and standard deviation of the MANIC(Gy) for the 6G / 1.36kg test condition were 0.41 and 0.15 respectively. Mean and standard deviation of the MANIC(Gy) for the 5G / 2 kg test condition were 0.31 and 0.08 respectively. Mean and standard deviation of the MANIC(Gy) for the PMHS tests were 0.80 and 0.65 respectively (see **Table 12**). Each of the data sets (both human subject as well as the PMHS) was analyzed for normality and all failed the Shapiro Wilk W test for normality in JMP. Thus non parametric tests were used to analyze the

distribution of MANIC(Gy) for males and females in each of the two human subject experiments.

**Table 10. 6G, 3 lb HMD Gy Experiment Human Subject Data Table**

| Subject | Mass (lb) | Gender | Crit Value (lb) | Peak MANIC(Gy) |
|---|---|---|---|---|
| Human 1 | 128 | F | 125 | 0.471 |
| Human 2 | 132 | F | 136 | 0.634 |
| Human 3 | 133 | F | 136 | 0.675 |
| Human 4 | 137 | F | 136 | 0.568 |
| Human 5 | 140 | F | 136 | 0.598 |
| Human 6 | 140 | F | 136 | 0.494 |
| Human 7 | 145 | F | 150 | 0.801 |
| Human 8 | 152 | M | 150 | 0.304 |
| Human 9 | 154 | F | 150 | 0.506 |
| Human 10 | 155 | M | 150 | 0.363 |
| Human 11 | 160 | M | 150 | 0.334 |
| Human 12 | 160 | M | 150 | 0.403 |
| Human 13 | 160 | F | 150 | 0.675 |
| Human 14 | 167 | F | 172 | 0.298 |
| Human 15 | 168 | M | 172 | 0.320 |
| Human 16 | 177 | M | 172 | 0.340 |
| Human 17 | 180 | M | 172 | 0.339 |
| Human 18 | 180 | M | 172 | 0.544 |
| Human 19 | 185 | M | 172 | 0.309 |
| Human 20 | 185 | M | 172 | 0.314 |
| Human 21 | 188 | M | 200 | 0.313 |
| Human 22 | 190 | M | 200 | 0.330 |
| Human 23 | 191 | M | 200 | 0.319 |
| Human 24 | 205 | M | 200 | 0.471 |
| Human 25 | 210 | M | 220 | 0.325 |
| Human 26 | 213 | M | 220 | 0.284 |
| Human 27 | 220 | M | 220 | 0.241 |
| Human 28 | 233 | M | 245 | 0.220 |
| Human 29 | 237 | M | 245 | 0.287 |
| Human 30 | 237 | M | 245 | 0.285 |
| Human 31 | 250 | M | 245 | 0.262 |

**Table 11. 5G, 4.5 lb HMD Gy Experiment Human Subject Data Table**

| Subject | Mass(lb) | Gender | Crit Values (lb) | Peak MANIC(Gy) |
|---|---|---|---|---|
| Human 1 | 132 | F | 136 | 0.359 |
| Human 2 | 135 | F | 136 | 0.447 |
| Human 3 | 140 | F | 136 | 0.527 |
| Human 4 | 140 | F | 136 | 0.383 |
| Human 5 | 140 | F | 136 | 0.337 |
| Human 6 | 155 | M | 150 | 0.298 |
| Human 7 | 155 | M | 150 | 0.257 |
| Human 8 | 156 | F | 150 | 0.409 |
| Human 9 | 160 | M | 150 | 0.260 |
| Human 10 | 160 | M | 150 | 0.385 |
| Human 11 | 164 | M | 172 | 0.232 |
| Human 12 | 165 | F | 172 | 0.258 |
| Human 13 | 167 | F | 172 | 0.424 |
| Human 14 | 170 | M | 172 | 0.227 |
| Human 15 | 180 | M | 172 | 0.272 |
| Human 16 | 180 | M | 172 | 0.313 |
| Human 17 | 188 | M | 200 | 0.257 |
| Human 18 | 190 | M | 200 | 0.310 |
| Human 19 | 193 | M | 200 | 0.278 |
| Human 20 | 204 | M | 200 | 0.255 |
| Human 21 | 206 | M | 200 | 0.241 |
| Human 22 | 210 | M | 220 | 0.269 |
| Human 23 | 215 | M | 220 | 0.253 |
| Human 24 | 227 | M | 220 | 0.265 |
| Human 25 | 230 | M | 220 | 0.232 |

For the 6G / 1.36kg test condition, females experienced a higher MANIC(Gy) compared to males (female mean=0.57, female median=0.58, male mean=0.33, male median=0.32, Independent-Samples Mann-Whitney U Test, p=0.0002). MANIC(Gy) values were negatively

correlated (to an $\alpha$=0.05) level with body mass (Spearman $\rho$ = -0.78, p<0.0001), sitting height (Spearman $\rho$=-0.57, p=0.0008), height (Spearman $\rho$=-0.67, p<0.0001), neck circumference (Spearman $\rho$=-0.73, p<0.0001), head circumference (Spearman $\rho$=-0.52, p=0.003), and age (Spearman $\rho$=-0.36, p=0.048).

For the 5 G / 2 kg test condition females experienced a higher MANIC(Gy) compared to males (female mean=0.39, female median=0.40, male mean=0.27, male median=0.26, Independent-Samples Mann-Whitney U Test, p=0.001). MANIC(Gy) values were negatively correlated (to an $\alpha$=0.05) level with body mass (Spearman $\rho$ = -0.595, p=0.0017), sitting height (Spearman $\rho$=-0.55, p=0.0045), height (Spearman $\rho$=-0.57, p=0.0031), neck circumference (Spearman $\rho$=-0.73, p<0.0001), and age (Spearman $\rho$=-0.41, p=0.04). MANIC(Gy) values were not significantly correlated with head circumference (Spearman $\rho$=-0.35, p=0.083).

Next, risk curves were constructed with SA using a combination of the aforementioned human subject and PMHS data to predict risk of AIS 2 or greater and AIS 3 or greater injury at a given MANIC(Gy) neck load. The PMHS experiment data table is provided in **Table 12**. The AIS 2 or greater risk function is shown in **Figure 21** and the AIS 3 or greater risk function is shown in **Figure 22**. For the AIS 2+ risk function data set, the MANIC(Gy) mean and standard deviation of the injurious data points was 1.07 and 0.71 respectively. The MANIC(Gy) mean and standard deviation of the AIS 2+ risk function non-injurious data points was 0.37 and 0.14 respectively. For the AIS 3+ risk function data set, the MANIC(Gy) mean and standard deviation of the injurious data points was 1.12 and 0.80 respectively. The MANIC(Gy) mean and standard deviation of the AIS 3+ risk function non-injurious data points was 0.38 and 0.15 respectively. The non-injury and injury data points are plotted at the location of their MANIC(Gy) values (x-axis) and at y-values of 0 or 100% respectively. Five data points were classified injurious at a level of AIS 2 or greater and 60 data points were non-injurious. A

comparison of the AIS 2+ risk curve and the AIS 3+ risk curve is provided in **Figure 23**. For the AIS 3 or greater risk function, four data points were injurious at a level of AIS 3 or greater and 61 data points were non-injurious. The difference between the AIS 2+ and AIS 3+ risk curves is produced by a single injury data point, indicating the sensitivity of the injury criteria when the PMHS injury data sample size is small, as it is in the current data set. Parr et al., experienced similar results in their development of AIS2+ and AIS 3+ risk curves to produce a modified Nij neck injury criteria for frontal impact (-Gx acceleration) using combined human subject and PMHS data (Parr et al., 2013).

**Table 12. Gy PMHS Data Table**

| Subject | Mass (lb) | Crit Value (lb) | Peak MANIC(Gy) | Acceleration (G) | AIS |
|---|---|---|---|---|---|
| PMHS 1 | 138.8 | 136 | 0.85 | 15.5 | 2 |
| PMHS 2 | 142.0 | 136 | 1.99 | 12.5 | 5 |
| PMHS 3 | 147.7 | 150 | 0.63 | 15.5 | 5 |
| PMHS 4 | 154.0 | 150 | 0.41 | 12.5 | 1 |
| PMHS 5 | 163.0 | 172 | 0.72 | 19.0 | 1 |
| PMHS 6 | 164.0 | 172 | 0.27 | 8.5 | 0 |
| PMHS 7 | 167.0 | 172 | 1.60 | 12.5 | 5 |
| PMHS 8 | 180.0 | 172 | 0.27 | 8.5 | 3 |
| PMHS 9 | 190.0 | 200 | 0.35 | 12.5 | 1 |

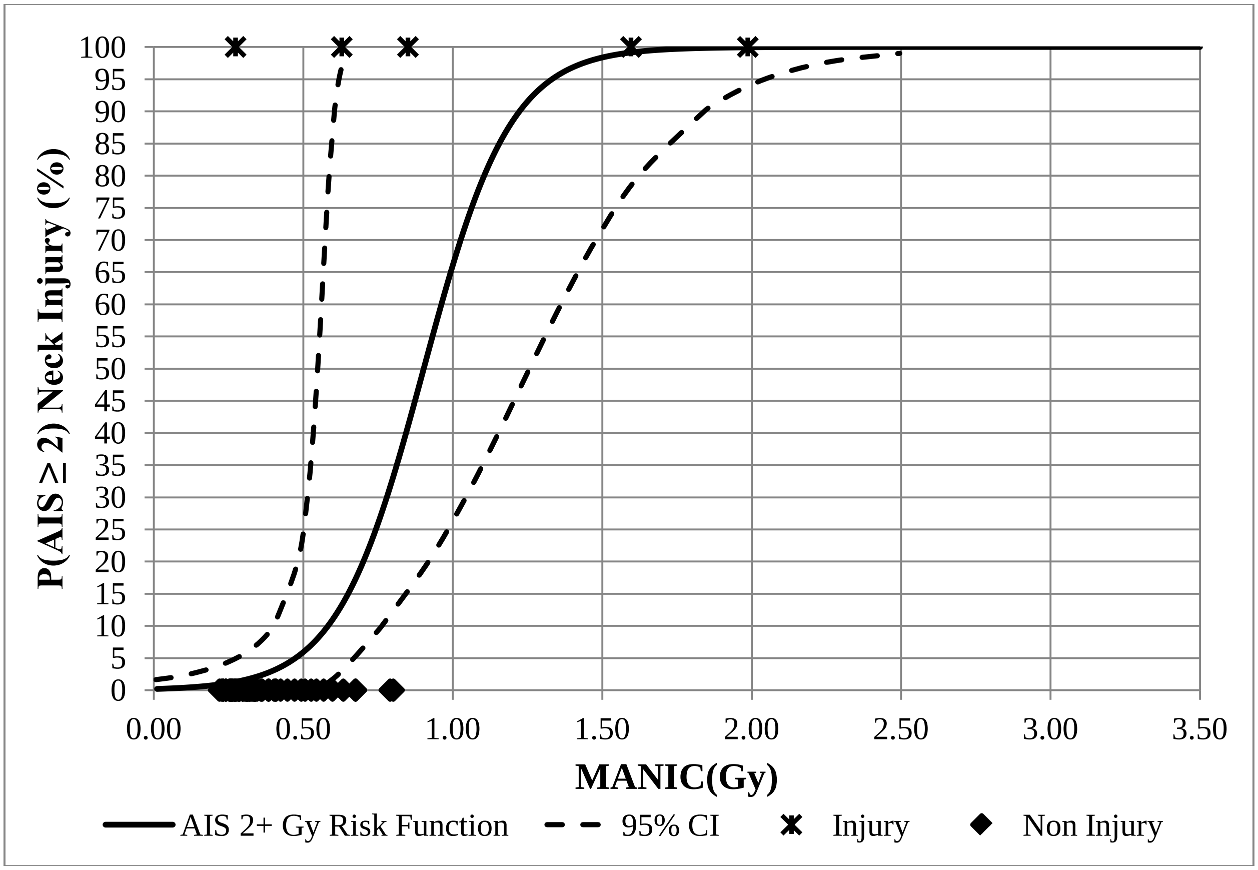


**Figure 21**. **Probability of AIS 2 or Greater MANIC(Gy) Risk Function with 95% CI**

The AIS 2+ risk function is provided in Equation 8.

$$P(AIS \geq 2) = \frac{1}{1 + e^{6.185 - 6.85 * MANIC(Gy)}} \quad (8)$$

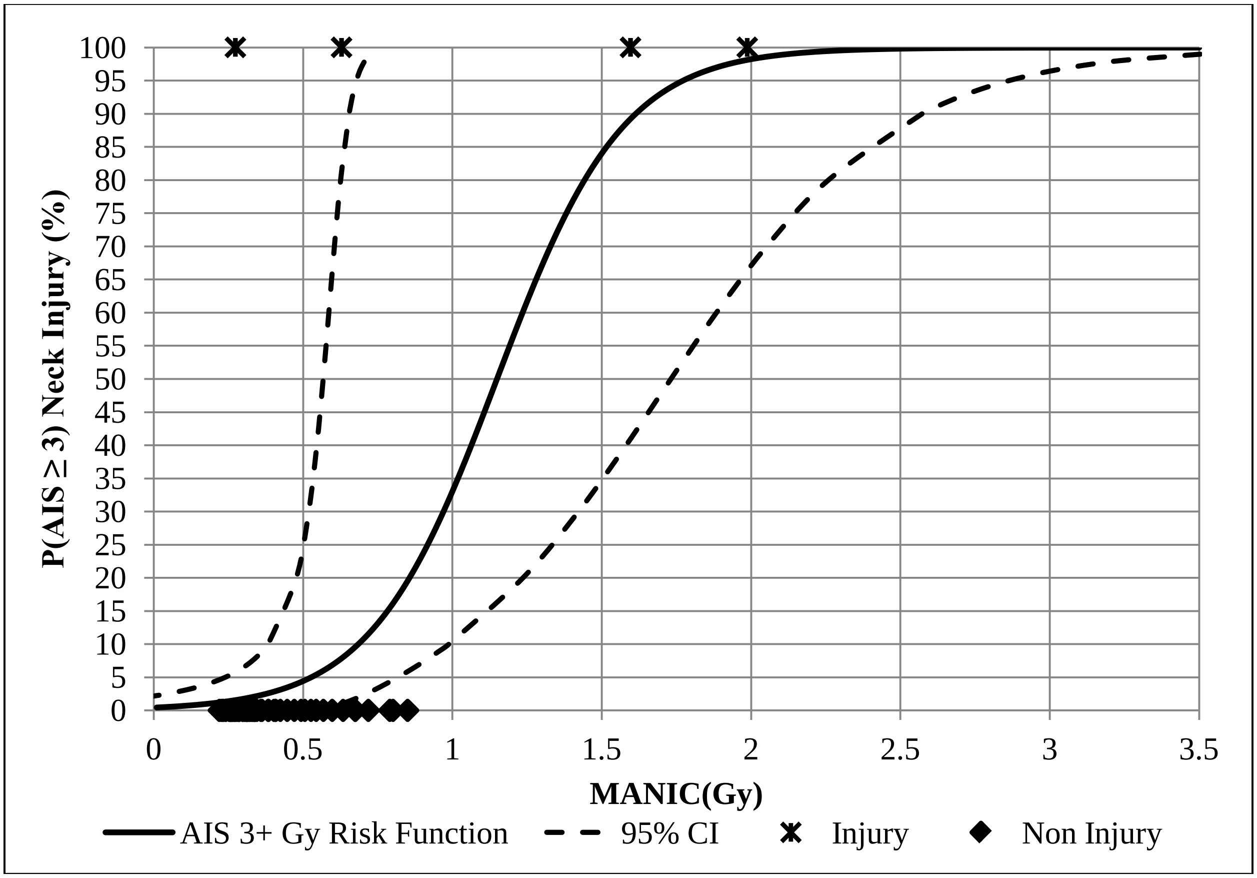


**Figure 22. Probability of AIS 3 or Greater MANIC(Gy) Risk Function with 95% CI**

The AIS3+ risk function is provided in Equation 9.

$$P(AIS \geq 3) = \frac{1}{1+e^{5.44-4.73*MANIC(Gy)}} \tag{9}$$

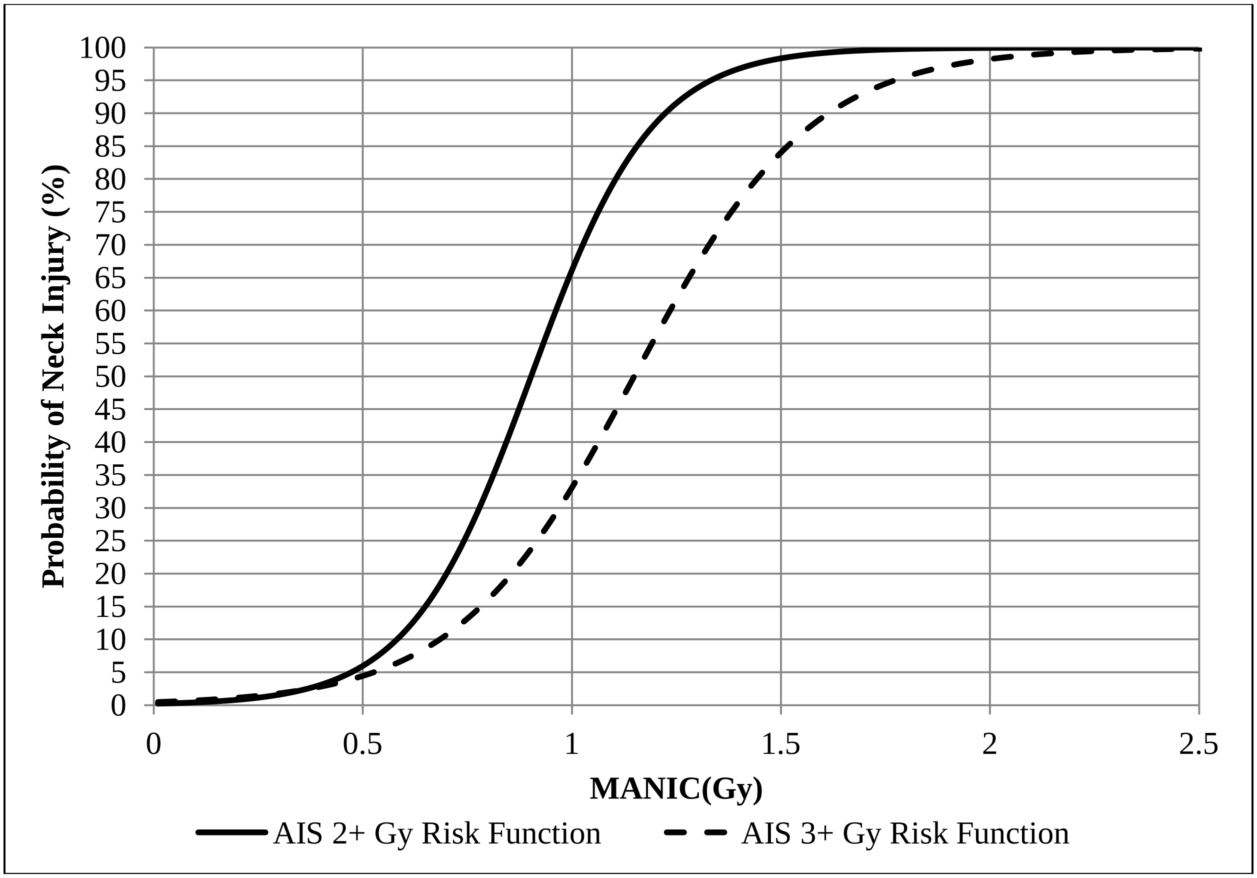


**Figure 23. AIS 2+ and 3+ Risk Functions**

The AIS+ and AIS 3+ risk functions in **Figure 23** behave as expected. A greater value for MANIC(Gy) is allowed at a specific risk level at the higher injury level. For example, at 5% risk of injury, the AIS 2+ risk curve allows for a MANIC(Gy) = 0.473 and the AIS 3+ risk curve allows for a MANIC(Gy) = 0.527 (see **Table 13** for 95% confidence intervals). Larger differences are observed at higher risk percentages as the two risk curves diverge between MANIC(Gy) values of 0.5 and 2.0.

## Discussion

The goal of this study was to analyze human subject upper neck data subject to Gy accelerative input with head supported mass using a multi-axial neck injury criterion formulation and then develop injury risk functions for the Gy axis of acceleration using combined human subject and PMHS data. The nature and availability of PMHS data has a large impact on injury risk function development. The small sample size of PMHS causes great sensitivity of the risk function to individual points. The AIS 3+ risk function differs from the AIS 2+ risk function by a single PMHS data point which had an observed AIS 2 neck injury which was considered an injury data point for the data set used to produce the AIS 2+ risk function but was classified in the non-injurious category for the data set used to produce the AIS 3+ risk function. Predicted MANIC(Gy) values and 95% confidence intervals are shown in **Table 13**. As expected, the AIS 3+ risk function allows for greater MANIC(Gy) values across all risk percentages. The difference in the 50$^{th}$ percentile upper and lower 95% confidence intervals for the AIS 2+ curve is 0.71 and 1.16 for the AIS3+ curve. The wider 95% confidence intervals observed might be attributed to the smaller injurious sample size of the AIS 3+ curve (N=4) compared to the AIS 2+ curve (N=5). Higher variability in MANIC(Gy) values was observed in the PMHS data than in the human subject data. It appears that the greater the accelerative input, the greater the variability in the data.

**Table 13. Predicted MANIC(Gy) Values (95% Confidence Intervals) at Various Risk Percentages**

| Risk Function | 5% | 10% | 20% | 50% |
|---|---|---|---|---|
| AIS 2+ | 0.473 (0.28, 0.67) | 0.58 (0.40, 0.76) | 0.70 (0.48, 0.92) | 0.90 (0.55, 1.26) |
| AIS 3+ | 0.53 (0.24, 0.82) | 0.68 (0.38, 0.99) | 0.86 (0.48, 1.23) | 1.15 (0.57, 1.73) |

The results of the intra-experiment statistical analysis provides insight into the behavior of the MANIC(Gy) structure and the critical values. As previously discussed, the purpose of the critical values is to normalize the neck loading to subject body mass as well as scale the relative importance of each individual load to the whole. If proper critical values are applied, it follows that MANIC(Gy) would not be correlated with subject anthropometric factors related to body mass. This was the case in analysis performed by Parr et al. with human subject experiments with frontal (-Gx) acceleration using NHTS's Nij neck injury criterion formulation, a combined loading structure, with a pair of thoroughly supported critical values (Parr et al., 2013). In this study, which involved a similar pairing of human and PMHS data, Nij was not correlated with body mass, sitting height, or neck circumference using Spearman's rank correlation.

In the current study for the 6G / 1.36 kg test, males and females experienced significantly different mean MANIC(Gy) values; additionally, correlations existed between MANIC(Gy) and body mass, head circumference, sitting height, height, head circumference, neck circumference, and age. For the 5G / 2 kg test, males and females experienced significantly different mean MANIC(Gy) values and correlations existed between MANIC(Gy) and body mass, sitting height, height, neck circumference, and age (head circumference was not correlated). Thus, the critical values are not effectively normalizing MANIC(Gy) to body mass and the other body mass related anthropometric properties. Instead, the presence of significant negative correlation indicates that these values are producing MANIC(Gy) values, which are larger than desired for smaller individuals.

### Example Application of MANIC(Gy)

The AIS 2+ risk function was applied to non-model building ATD data sets from the AFRL Biodynamics data base to demonstrate how the MANIC(Gy) could be applied to assess

the risk associated with varying helmet mass and accelerative input. This ATD data was collected in the same overall experiment that generated the human subject data used in this paper. To evaluate the effect of HMD mass, a 5G / 0lb HMD test was compared to a 5G / 4.5lb HMD test (acceleration held constant). Three large ATDs with Hybrid-III necks were tested in both test configurations. The MANIC(Gy) was computed over the time history, the peak MANIC(Gy) was identified, then the results from the two different test configurations were compared. First, a statistical comparison for significant differences in MANIC(Gy) between the two tests was performed. Then the difference in risk posed by the 4.5lb HMD versus no HMD was compared. The 4.5 lb HMD difference under same acceleration resulted in no significant difference in MANIC(Gy) (Mann-Whitney U test, $p=0.83$, $\alpha=0.05$).

To evaluate the effect of accelerative input, a 4G / 3lb HMD test was compared to a 6G / 3 lb HMD test (head supported mass held constant). Four large ATDs with Hybrid-III necks were tested in both configurations. With a constant 3lb HMD and a change in accelerative input from 4 to 6 G a significant difference was observed in mean peak MANIC(Gy) (Mann-Whitney U test, $p=0.02$, $\alpha=0.05$). The 4G / 3lb HMD mean peak MANIC(Gy) was 0.11 (median=0.11) and the 6G / 3lb mean peak MANIC(Gy) was 0.22 (median=0.22). These peak mean and median MANIC(Gy) values represent a 0.44% and a 0.92% risk of AIS 2+ neck injury, respectively. The addition of 2 Gs of acceleration resulted in approximately twice the risk of AIS2 + neck injury.

This study was limited by the availability of adequate data to fully support the criteria. The formulation presented here excluded $Mx$ in MANIC(Gy) as this component was missing from the human subject data available to the authors. To assess the potential impact of this on the final criterion, the PMHS data which included the side bending moment ($Mx$) component was analyzed. Of the nine PMHS data points, the mean peak MANIC(Gy) (with no $Mx$) was 0.79

(median=0.63), whereas the mean peak MANIC (including *Mx*) was 0.86 (median=0.74). A statistical comparison showed the means were not statistically different, thus based on this small sample size statistical test, it would appear that the MANIC(Gy) sub model can be reasonably used as an adequate surrogate for the full model (Mann-Whitney U test, p=0.45, α=0.05). However, to allow for the development of a complete six load lateral impact multi-axial neck injury criterion, future human subject experiments should observe all six primary upper neck loads. This should be possible with the current state of technology where channels for data processing and sensor size are no longer constraining factors. When human subject data becomes available in the future with all six primary neck loads, the criterion developed in this work should be updated to include the full MANIC formulation shown previously in Equation 6.

However, the critical values proposed by Nichols for *Mx* should also be considered carefully. Yoganandan et al. established a reference value for side bending at 75 N-m (Yoganandan et al., 2011). Their definition of reference value is that the 24 PMHS tests in their experiment showed tolerance of 75 N-m of coronal moment without failure. In the PMHS data set *Mx* values had a mean of 51.2 N-m (min of 13.4 and max of 85.1 N-m) and a standard deviation of 21.6 N-m. The critical values proposed for use by Nichols range from 67 N-m for the smallest subject in the eight category critical value matrix to 209 N-m for the largest subjects (see **Table 9**), where the value for the largest subjects exceed the reference value proposed by Yoganandan by a considerable margin.

The current study was also limited by the application of combined, existent human and PMHS data. First, there were differences in test set up between the human and PMHS tests. The human subjects were seated in a representative ejection seat, while the PMHS were tested using one of three variations of a side-facing cargo aircraft seat and associated restraints. This is a potential source of error in the observed neck response data. Second, human subjects and PMHS

behave differently in accelerative testing due to the difference between active and passive musculature. Third, the human subjects and PMHSs had different means for both body mass and age (mean human subject body mass was 80.0 kg / 176.3 lb, mean human subject age was 30.8 yr, mean PMHS body mass was 72.9 kg / 160.7 lb, mean PMHS age was 55.9 yr), potentially affecting the predictive ability of the risk function to human subjects at higher MANIC(Gy) values. Using PMHS neck load and injury data to construct the risk curve provides a more conservative criterion. This is because PMHS, which typically have less body mass, greater age, and lack active musculature (which translates into higher neck loads), typically experience injury at less input load than human subjects. Thus a criterion that establishes a load limit from a risk function constructed with PMHS data would typically be shifted left compared to a risk function constructed with human subject data. This results in less allowable load at any percentage risk of injury, which is desirable from a pure safety perspective. However, a criterion that is too conservative may prevent systems from incorporating other important capabilities. These tradeoffs between system safety and system capability must be considered in the domain-specific application of the risk function development methods proposed in this work.

Application of the human subject risk function to evaluate side impact accelerative tests requires the assumption be made that the ATD is structurally and kinematically biofidelic. This assumption has been challenged by the results of some research, especially with the use of head supported mass (Bass et al., 2006; Salzar et al., 2009). It is recommended that future work attempt to develop transfer functions to more appropriately apply this human risk criterion using results from accelerative testing from the Hybrid-III ATD neck. It is unlikely that a different ATD neck will be used in the DoD escape system qualification testing protocol any time in the near future. While the development of an improved, more biofidelic ATD neck is highly desirable, the use of the Hybrid III neck is entrenched in established testing protocols within the

DoD and automotive industry, necessitating transfer functions if warranted by the investigations of future research.

It is proposed that the AIS 2 + risk function constructed in this paper be used as a basis for a preliminary side impact risk criterion MANIC(Gy) limit of 0.48. Systems performing below this level would limit pilot risk to a 5% or less probability of AIS 2 or greater injury when subjected to Gy loading. Below 0.48 would be considered acceptable risk according to the AF escape community, above would be considered unacceptable risk. However, it should be noted that because a risk function exists, decision makers have the ability to make trade decisions to accept higher risk of injury if desired based upon the system cost and schedule implications of system modification to reduce risk. Further research is warranted to improve the MANIC(Gy) criteria, specifically, the current research illustrates the need to develop improved critical values for this criteria.

# VI. Development of an Updated Tensile Neck Injury Criterion

## Chapter Overview

The paper that comprises this chapter has been accepted for publication in the Journal of Aviation, Space, and Environmental Medicine. This paper outlines the development of an updated AF tensile neck injury criterion. In this paper, the risk functions for the -Gz axis of acceleration (**Figure 24**), which results primarily in a tensile loading of the pilot's neck, is constructed. No adequate PMHS neck load and injury classification data were available in the literature from +Gz accelerative experiments. Human subject data from experiments in the +Gz axis were available, but risk function development, as outlined in the methodology, require both human subject and PMHS data. However, adequate PMHS tensile (-Gz) loading and injury classification data and human subject tensile loading data were available in the literature. Therefore, a tensile risk function was constructed as the basis for the Gz portion of the multi-axial neck injury criteria – MANIC(Gz). Thus, the tensile neck injury criteria developed in this chapter is used as the z-axis portion of the overall multi-axial neck injury criteria.

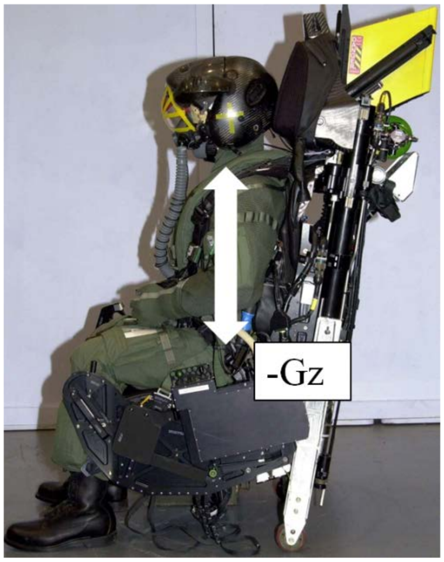


**Figure 24. -Gz Axis of Acceleration**

**Abstract**

**BACKGROUND:** Ejection neck safety remains a concern in military aviation with the growing use of helmet mounted displays (HMDs) worn for entire mission durations. The original USAF tensile neck injury criterion proposed by Carter et al. (Carter et al., 2000) is updated and an injury protection limit for tensile loading is presented to evaluate escape system and HMD safety. **METHODS:** An existent tensile neck injury criterion was updated through the addition of newer post mortem human subject (PMHS) tensile loading and injury data and the application of Survival Analysis to account for censoring in this data. The updated risk function was constructed with a combined human subject ($N$ = 208) and PMHS ($N$ = 22) data set. **RESULTS:** An updated AIS 3+ tensile neck injury criterion is proposed based upon human and PMHS data. This limit is significantly more conservative than the criterion proposed by Carter et al. (Carter et al., 2000), yielding a 5% risk of AIS 3+ injury at a force of 1136 N as compared to a corresponding force of 1559 N. **DISCUSSION:** The inclusion of recent PMHS data into the original tensile neck injury criterion results in an injury protection limit that is significantly more conservative, as recent PMHS data is substantially less censored than the PHMS data included in the earlier criterion. The updated tensile risk function developed in this work is consistent with the tensile risk function published by the Federal Aviation Administration used as the basis for their neck injury criterion for side facing aircraft seats.

**Keywords:** HMD; pilot; ejection safety; risk curves

## Introduction

Ejection neck safety remains a concern in military aviation with the growing use of helmet mounted displays (HMDs) worn for entire mission durations. The development of neck injury risk functions and criteria for all axes of acceleration are important to adequately develop and test military aircraft HMDs and escape systems to provide for pilot safety (Stemper et al., 2009). Previous studies have developed aviation specific risk curves for -Gx acceleration (Bass et al., 2006; Parr et al., 2013) and have proposed ejection neck injury criteria adapted from automotive criteria (Nichols, 2006). However, neck injury risk criteria developed specifically for the ejection environment for the other primary axes of acceleration are also important as the pilot is exposed to dynamic tensile neck loads during the windblast and parachute opening phases of ejection (Carter et al., 2000). This work focuses on the -Gz axis of acceleration which results in a primary neck load of axial tension. Additionally, tensile loading in combination with neck flexion or extension moments have been observed throughout the ejection sequence (Pellettiere et al., 2005). The purpose of this paper is to develop and propose an updated United States Air Force (USAF) tensile neck injury criterion and to compare this tensile criterion to legacy criteria.

Ejection safety criteria guidelines have been established by the USAF escape system oversight office, including a 5% risk of Abbreviated Injury Scale (AIS) 2 or greater neck injury (AAM, 2008; Parr et al., 2013). It is desirable that new or updated injury risk criteria adhere to this guidance. However, a baseline injury threshold of AIS 3 or greater was used in both the original tensile risk criterion (Carter et al., 2000) and also in a more recent cadaver tensile neck strength experiment (Yliniemi et al., 2009) that, to the authors’ knowledge, provides the most comprehensive, post mortem human subject (PMHS) tensile neck injury data set available in the

literature. Thus the available data restricts the development of an updated tensile neck injury criterion to injury levels of AIS 3 or greater rather than AIS 2 or greater.

The method of statistical analysis chosen to construct an injury risk function is important in order to obtain an accurate risk criterion. Survival analysis (SA) is a method of statistical analysis which accounts for censored data which is common in the field of injury biomechanics (Hosmer et al., 2008). Survival Analysis is emerging as the accepted standard for generating injury risk functions when human subject and PMHS data are used (Bass et al., 2006; Cutcliffe et al., 2012). Human subjects are not loaded to the point of injury, thus human subject data is right censored as the load at which injury is likely to occur is to the right of (greater than) the observed load value. On the other hand, since the exact load at which injury occurs is usually unknown without the use of special equipment, PMHS data is often left censored as the load value at which injury occurs is likely to be to the left (less than) of the observed load value. Standard logistic regression (LR) statistical methods assume exact data is being used. Widely used neck injury risk functions have been constructed using LR rather than SA, including the National Highway Transportation Safety Administration's frontal impact neck injury criteria (called the Nij) and the Federal Aviation Administration's (FAA) neck injury criteria for side facing aircraft seats (Eppinger et al., 2000; FAA, 2011). At the time these risk functions were created, LR was the dominant and preferred statistical method. While the original tensile neck injury criterion (Carter et al., 2000) was constructed using standard LR, the present study recreated it using SA and will primarily use SA rather than standard LR to develop and propose an updated tensile neck injury criterion. Other improvements to LR are also available which makes the continued use of LR with injury biomechanics data more appropriate. Firth's Adjusted Maximum Likelihood method of LR, which uses the generalized linear model, reduces bias in the parameters due to low subject counts or when data is skewed toward one outcome, and accounts

for missing or limited data and will be incorporated in the present analysis (Firth, 1993). Firth's method is used to generate risk functions for comparative purposes in this paper when SA is not a viable option due to complete separation between the human subject and PMHS data (lack of one or more overlapping injury/non-injury data points).

Various methods of modern risk function construction have been outlined in other work, to include using data from matched pair anthropometric test device (ATD) and PMHS experiments (FAA, 2011), combined human subject and PMHS data (Carter et al., 2000; Parr et al., 2013; Pellettiere, 2012), matched pair ATD and piglet experiments scaled to human applicability (Eppinger et al., 1999), and instrumented PMHS neck section (Bass et al., 2006). The current research employs the method of combining human subject and PMHS data to propose an updated AF tensile neck injury criterion. The updated criterion will follow the method to construct the original risk curve (Carter et al., 2000). The main benefits of this method are that accurate human neck loading and injury are observed and that these observations are incorporated directly into the risk function. These two entities are arguably the most important elements of an accurate and applicable risk function. There are also some limitations to this method of risk function development. In the field of injury biomechanics, experiments are often expensive and data are difficult to collect. Neck response to non-injurious loading can be collected from human subject testing, but the loading is estimated from observed head accelerations combined with subject anthropometry (Parr et al., 2013). Neck response and injury data collected from experiments with PMHS may not be representative of the typically young, fit military flying population and lacks active musculature, potentially resulting in overly conservative criteria. Human subject testing requires extensive approval procedures from Institutional Review Boards, and PMHS testing is limited to available specimens that require careful storage, handling, and injury assessment procedures (e.g. necropsy by trained personnel

and radiographic scans). These constraints typically result in small sample sized for human and PMHS experiments.

From the standpoint of statistical integrity, data from a single controlled experiment is the most desirable for risk function construction. In the case of human risk function generation, injurious experiments are not performed on human subjects and PMHS testing is limited to available specimens that require specialized care and procedure, thus necessitating a combination of human subject and PMHS data sets for risk function estimation. In this instance, it is desirable to minimize the number of data sets. Ideally, data from a single human subject experiment and a single matched PMHS experiment with conditions controlled closely for error reduction and uniformity between the two is desired, the primary difference being accelerative input. However, to constitute an adequate PMHS sample size and achieve improved statistical model inference ability, this is not always possible. When multiple PMHS neck load data sets are used to generate a single risk function, the experimental conditions are evaluated to ensure adequate similarity and lack of bias. Additionally, comparison of the observed load distributions using the appropriate parametric or non-parametric statistical tests is important to ensure that it is reasonable to combine the PMHS data sets.

The original tensile risk function employed a novel technique for risk curve construction by using a combination of human subject and PMHS data in the regression (Carter et al., 2000). The PMHS data used to construct the original curve came from three separate studies consisting of six, three, and one subject for a total of 10 PMHS tensile data points. Of the 10, three were non-injurious and seven were injurious at an AIS level of 3 or greater. The study producing six data points was a whole cadaver frontal impact study where a combination of tension and flexion occurred in the subjects and injury was observed by post test autopsy (Cheng et al., 1982). Cervical spine tension of PMHS was calculated using observed accelerations and head/neck

mass properties. Three of the data points came from a study where pure axial tension was applied to the cervical spine of whole cadavers; loads were recorded and injury was specified (Yoganandan et al., 1996). The study that produced the remaining single data point was the only isolated spinal column tested and no injury was observed (Sances et al., 1981). Post mortem human subjects are expensive to use in experimentation and are available in limited numbers. Thus, this method of combining PMHS data from pure tension studies and combined loading studies has the benefit of increasing total subject count to improve risk curve estimation. However, it has been observed in other studies that the neck is more susceptible to injury when subject to combined bending (forward flexion, rearward extension) and tensile loading as compared to pure tensile loading, thereby potentially affecting the injury prediction of the risk function (Eppinger et al., 1999; FAA, 2011).

The original risk curve scaled the PMHS tension values by a factor of 1.5 in order to account for age (20%) and use of PMHS (25%) (Carter et al., 2000). Depending on the application, this scaling may be desirable. However, a recently developed FAA risk criteria for side facing aircraft seats (FAA, 2011) was constructed using the unscaled tensile loads of the EuroSid-2 (ES-2) ATD corresponding to the neck response of similarly loaded PMHSs. Additionally, Parr et al. and Bass et al. have proposed frontal impact neck injury criteria formulations using unscaled PMHS loads (Bass et al., 2006; Parr et al., 2013). In the aviation environment pilot safety is of utmost importance. Therefore, this paper will analyze risk curves constructed using the more conservative, unscaled tensile loads for use as an updated AF tensile neck injury criterion. If the practitioner sees value applying scaling factors to their data it can be incorporated into their domain-specific injury risk function construction methodology.

## Methods

### *Subjects*

The original curve used human subject experimental neck load data from previous Air Force Research Laboratory (AFRL) tests where the subject was seated in a test ejection seat, but oriented horizontally (Brinkley and Getschow, 1988). Acceleration was applied to the seat such that the subject's body was accelerated away from their head, resulting in a neck response that was observed to be primarily tensile loading of the cervical spine. No new human subject experiments in this orientation have been completed since the original tensile criterion was published; therefore this data remains the best source of human subject neck tension data and thus was used to create the updated risk functions. This AFRL data set was used as the human subject data set in each of the risk function constructed in this paper.

The original tensile risk curve paper recommended that future work should add any newly collected PMHS neck tension data to the risk function when it came available to increase the statistical power of the risk function (Carter et al., 2000). In the time since the original tensile neck injury risk criterion was created, a PMHS study specifically developed for the purpose of providing data to improve the original risk curve was published (Yliniemi et al., 2009). Yliniemi et al. conducted tensile load testing to failure on 12 PMHS head and torso specimens where the skin and musculature were intact and T8-T11 were potted and secured to the base of the loading apparatus (Yliniemi et al., 2009). Eight males and four females were tested using aviation specific tensile loading rates ranging from 520 mm/s to 740 mm/s (Yliniemi et al., 2009). Mean subject anthropometry included the following: age (50.1 yrs), height (173.5 cm), and body mass (76.7 kg). Failure loads were recorded as well as detailed cervical spine injury from post-test radiographs using current AIS values. All subjects experienced AIS 3 or

greater neck injury, and the mean tensile load at failure was 3100 N (3250 N for males, 2803 N for females).

**Table 14** provides a summary of the failure loads and anthropometry of all PMHS.

**Table 14. PMHS Peak Tensile Neck Load and Anthropometry**

| Sex | Type | Age (year) | Body Mass (kg) | Failure Load (N) | AIS 3+ Injury | Source |
|---|---|---|---|---|---|---|
| **M** | Whole | 66 | 72.5 | 3490 | Yes | Cheng et al., 1982 |
| **F** | Whole | 54 | 50 | 7200 | Yes | Cheng et al., 1982 |
| **M** | Whole | 56 | 96 | 2420 | Yes | Cheng et al., 1982 |
| **M** | Whole | 63 | 72.5 | 850 | No | Cheng et al., 1982 |
| **M** | Whole | 68 | 93 | 6520 | Yes | Cheng et al., 1982 |
| **M** | Whole | 67 | 60 | 3210 | No | Cheng et al., 1982 |
| **N/A** | Whole | 66±11 | N/A | 2400 | Yes | Yoganandan, 1996 |
| **N/A** | Whole | 66±11 | N/A | 3900 | Yes | Yoganandan, 1996 |
| **N/A** | Whole | 67 | 67 | 3800 | Yes | Yoganandan, 1996 |
| **N/A** | Isolated | 61 | 70 | 2688 | No | Sances, 1981 |
| **F** | Torso | 48 | 55 | 3560 | Yes | Yliniemi et al., 2009 |
| **F** | Torso | 45 | 59 | 2250 | Yes | Yliniemi et al., 2009 |
| **F** | Torso | 56 | 68 | 1910 | Yes | Yliniemi et al., 2009 |
| **F** | Torso | 43 | 74 | 3490 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 35 | 59 | 4060 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 48 | 73 | 3860 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 50 | 68 | 2810 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 60 | 77 | 3150 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 59 | 82 | 3230 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 37 | 77 | 3220 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 59 | 114 | 2440 | Yes | Yliniemi et al., 2009 |
| **M** | Torso | 61 | 114 | 3230 | Yes | Yliniemi et al., 2009 |

AIS = Abbreviated Injury Scale

*Procedures*

The original tensile risk function from Carter et al. was reconstructed with the combined human subject and unscaled PMHS source data using LR. The exact estimates were successfully recreated. From this baseline, risk functions were constructed with both SA and LR using Firth's method in addition to observing various combinations of the PMHS source data. The curves were then compared to determine which was best suited for application as an updated tensile neck injury criterion for the aviation environment. "SA" refers to survival analysis whereas "Firth's" denotes Firth's Adjusted Maximum Likelihood method of LR. Firth's adjustment is used when LR coefficients might be biased due to data being skewed toward one outcome. In the case of the present data set with 208 human subject data points compared with between 10 and 22 PMHS data points, this bias reduction method is appropriate. It was also investigated whether or not constructing separate risk functions for large and small individuals was supported by the data.

*Statistical Analysis*

Risk functions were initially constructed using SA (Hosmer et al., 2008) after methods used in research by Bass et al. and Parr et al. (Bass et al., 2006; Parr et al., 2013) and then also with LR using Firth's method. The USAF escape community is interested in limiting injury risk at the 5% level, thus risk functions were evaluated based upon the 5% predicted tensile load and 95% confidence intervals at a 5% risk of AIS 3 or greater neck injury. Risk curves produced with the 'Original' data (ORG) and the 'Combined' data (COM) using SA were compared. 'Original' is the risk function constructed using the PMHS data from the original criteria (N = 10) (Carter et al., 2000). 'New Only' (NEW) is a risk function built using only the new data from the Yliniemi et al. study (N = 12) (Yliniemi et al., 2009). 'Combined' is a risk function

built using a combination of the original risk function PMHS data combined with the new PMHS data (N = 22). There were no significant differences between ORG tensile failure PMHS data and the NEW tensile failure data (Mann-Whitney, $\alpha = 0.05$, $P = 0.717$), suggesting that combining these two data sets would not be inappropriate. Risk functions generated with the ORG and COM PMHS data using SA were compared to investigate the effects of adding the new data to the original curve. Additionally, to examine the difference between the ORG, NEW, COM risk functions, LR with Firth's method was applied as it is the only method to generate parameter estimates for the NEW risk function since this data includes no noninjurious loads.

## Results

The AIS 3 or greater risk function generated with the COM data is compared to the ORG unscaled risk function in **Figure 25**. The 5% predicted tensile loads are 1136 N and 1559 N respectively. This reduction in predicted load at 5% risk of AIS 3+ injury is a result of the nature of the COM data compared to the ORG data. The ORG PMHS data ($N = 10$) included a majority of data points ($N = 6$) from highly accelerative loading (32-39 G) resulting in a greater average neck tensile loading (mean of 3648 N). The NEW data ($N = 12$) was generated by experimental conditions that incorporated aviation loading rates which generated lower injurious tensile neck loads (mean of 3100 N), adding fidelity to the risk function at the load ranges applicable to aviation. **Table 15** provides a summary of the results from the regression analysis.

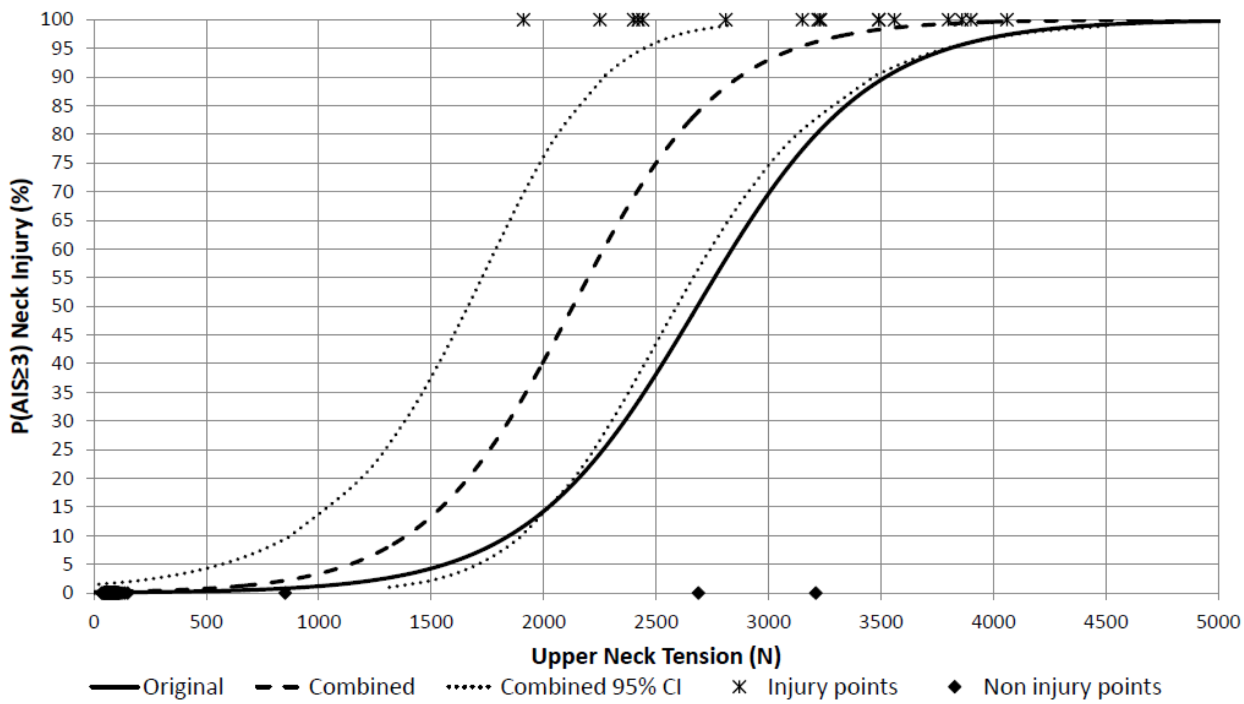


**Figure 25. Probability of Abbreviated Injury Scale (AIS) 3 or Greater 'Original' and 'Combined' Risk Functions Constructed Using Survival Analysis (95% CI of Combined Risk Curve Shown)**

**Table 15. 5% Risk of Abbreviated Injury Scale 3+ Injury Predicted Tension and 95% Confidence Intervals (Upper, Lower), All Un-Scaled Values**

| PMHS Data Source | Regression Method | |
|---|---|---|
| | SA | Firth's |
| **Original (N)** | 1559 (756, 2358) | 1349 (415, 2016) |
| **New Only (N)** | N/A | 864 (211, 1516) |
| **Combined (N)** | 1136 (566, 1706) | 1013 (344, 1472) |

PMHS = Post Mortem Human Subject; SA = Survival Analysis; Firth's = Firth's Adjusted Maximum Likelihood Method of Logistic Regression; N/A = no parameter estimates due to separation in the data

In addition to their originally proposed risk function, Carter et al. also presented an alternate curve applicable to lower body mass individuals based upon the addition of smaller individuals to the pilot population and the recent addition of female pilots in ejection aircraft (Carter et al., 2000). This was achieved by scaling the data down by 25% to generate a curve for smaller individuals (<73kg) in addition to the curve for larger individuals (>73 kg) (Carter et al., 2000). Scaling was used as a theoretical adjustment since the sample size of available PMHS data was too small to draw any statistical conclusions based upon body mass or gender. At the outset of the present study it was hoped that the additional 12 PMHS data points would allow for two separate curves to be generated based upon actual mass or gender data of the combined 22 data points. To analyze the effect of body mass, peak tension of subjects with mass greater than the sample mean of 76.7 kg ($N = 13$) was compared with peak tension of subjects with mass less than 76.7 kg ($N = 5$). Body mass was not reported for two subjects. To analyze the effect of gender, peak tension of female subjects ($N = 5$) was compared with peak tension of male subjects ($N = 13$). Gender was not reported for four subjects. Based on this data set, neither body mass nor gender was a significant predictor of tensile neck loading using the Mann-Whitney U non-parametric test for body mass ($\alpha= 0.05$, $P = 0.72$) and gender ($\alpha =0.05$, $P = 0.84$). Therefore, the data was pooled and a single risk function for all body masses was created.

A comparison was made between the ORG unscaled risk function, the NEW risk function, and the COM risk function. Both standard LR and SA failed to produce parameter estimates for the NEW data set, thus LR using Firth's method was used to generate all three functions for the purposes of comparison (**Figure 26**). It should be noted that these curves, generated with Firth's method of LR, were generated for comparison purposes only and that the curve depicted in **Figure 25**, generated with SA, will be used as the basis for the updated USAF tensile neck injury criterion. The NEW risk curve resulted in the most conservative tensile

values at all percentages of AIS 3+ neck injury, followed by the COM risk curve. The ORG allowed for the highest tensile loads. The Firth's method 5% predicted risk of AIS 3+ neck injury for the NEW, the COM, and the ORG unscaled curve were 864, 1013, and 1349 N respectively and fall within the confidence bounds generated in **Figure 25**.

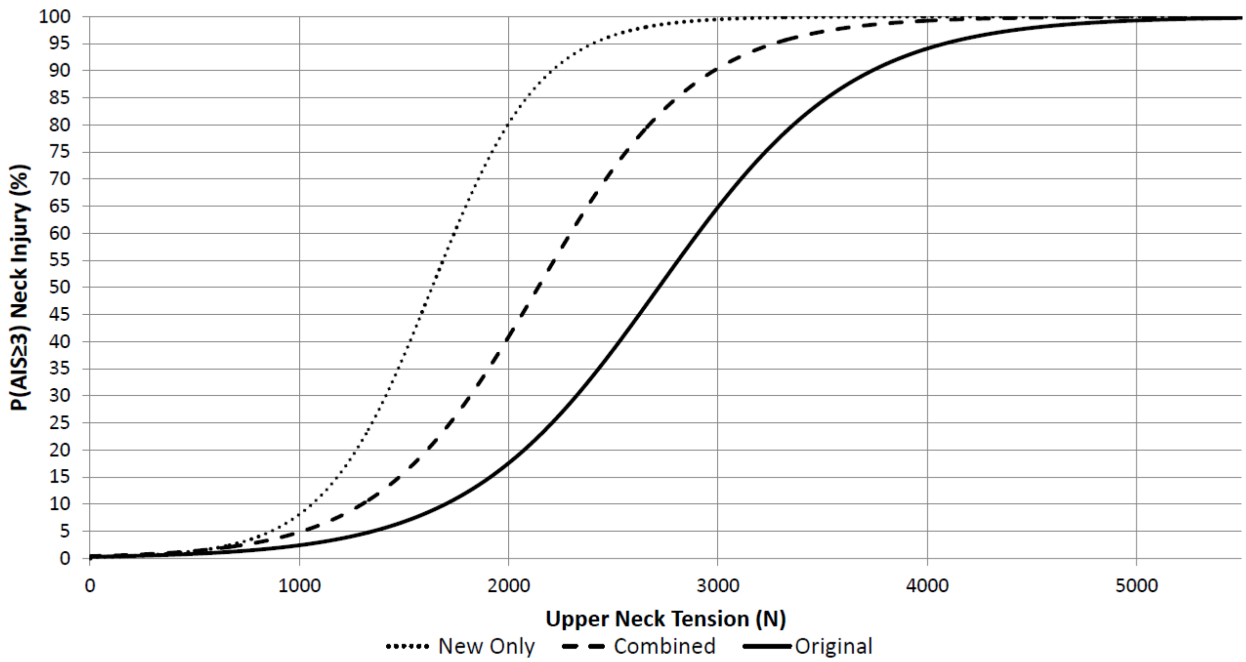


**Figure 26. Probability of Abbreviated Injury Scale (AIS) 3 or Greater 'Original' Risk Function Compared to Risk Functions Constructed with 'New Only' Post Mortem Human Subject (PMHS) Data and with the 'Combined' PMHS Data Constructed Using Firth's Method of Logistic Regression**

## Discussion

As seen from **Figure 25**, the COM risk function is more conservative at all risk levels than the ORG. This follows logically based upon the addition of 12 injurious data points with a lower mean (3100 N) than that of the data used to construct the ORG curve (4247 N). It is

possible the Cheng et al. data is extremely left censored due to the significantly higher input acceleration and that these six values artificially inflate the overall risk curve for allowable tensile loading (Cheng et al., 1982). Since the Cheng et al. data remains the second largest known whole PMHS data set, it cannot be excluded from the analysis. The 12 Yliniemi et al. data points, while not exact, are likely to be less left censored since failure was defined by observing decreasing load in conjunction with increasing displacement on a load-distraction curve (Yliniemi et al., 2009). The Yliniemi et al. study was set up to carefully observe and control the single axis load application and recorded injury loads with less error (SD=645 N) compared to the dynamic, combined loading Cheng et al. study (SD=2455 N) (Cheng et al., 1982; Yliniemi et al., 2009).

Using Firth's method of LR to compare the NEW, the COM, and the ORG risk functions, the resulting risk curves behaved as expected. The NEW was the most conservative, followed by the COM curve, and the ORG curve, which allowed for the most tension at all risk percentage levels. The difference in the 5% predicted risk of AIS 3+ injury observed between the ORG unscaled curve and the NEW curve can be explained partly by understanding the underlying data used to create the two curves, specifically the PMHS data used. The mean of the AIS 3+ injurious PMHS tensile load values used in the unscaled ORG curve was 4247 N. The mean of the AIS 3+ injurious tensile loads from the Ylinimi et al. study was 3100 N (Yliniemi et al., 2009). This difference causes the regression of the Ylinimi et al. data set paired with the same non-injurious human subject data as the ORG curve, to predict injury at lower input levels of tension.

The results of the COM risk function of this study compare well with the FAA neck injury criteria for side facing aircraft seats (FAA, 2011) developed over the course of nine years through an extensive collaborative research effort. The FAA injury criterion is the most recently

developed and robust of its kind, incorporating state of the art PMHS and ATD testing performed at premiere injury biomechanics research institutions. The researchers that developed the FAA side impact criteria found that only tensile loading was predictive of injury in the PMHS tested (FAA, 2011). Thus, similar to the criterion developed in this work, the FAA criterion is also a tensile-only loading criterion, though the FAA criterion assumes that some shear force or bending torque is present as a necessary condition for the tensile loading to cause injury. The FAA tensile criterion research effort culminated with the publication of the final risk function and criterion, constructed with data collected from 10 matched pair PMHSs and ES-2 ATDs at various accelerative levels and side-facing aircraft seat configurations (FAA, 2011). It should be noted that these 10 PMHS tests are separate from the data used to construct the tensile risk function in this paper; the FAA criterion data points and associated risk function are used as a validation data set for the updated USAF tensile criterion proposed in this study. The FAA risk criterion is used to evaluate and qualify new side facing aircraft seats for use in commercial aviation in a similar fashion as the updated USAF tensile neck injury criterion would be used to evaluate and qualify new HMDs and escape systems. **Table 16** compares the results of the proposed US AF updated tensile neck injury criteria with the applicable predicted values of the FAA criteria.

**Table 16. Comparison of Federal Aviation Administration Neck Injury Criteria for Side-Facing Aircraft Seats with Combined Risk Function from Present Study**

| Risk Function | Predicted Value |
|---|---|
| **Combined AIS 3+ 25% (90% CI)** | 1758 (1370, 2150) N |
| **FAA AIS 3+ 25% (no 90% CI given)** | 1800 N |
| **Combined AIS 3+ 50% (90% CI)** | 2128 (1738, 2517) N |
| **FAA AIS 3+ 50% (90% CI)** | 2308 (1755, 2861) N |

AIS = Abbreviated Injury Scale

The COM risk function developed here is within 3% of the FAA tensile criteria at the 25% AIS3+ injury risk level and within 8% at the 50% AIS3+ injury risk level. The FAA criteria only reported the 25% and 50% values and the curves could not be recreated with the data presented in the report, thus the 5% values could not be compared. While the primary loading mechanism of the FAA criteria was side impact, it is interesting that the two criteria arrive at similar risk functions for cervical spine tensile loading. While the current study's COM risk function was primarily constructed with pure tensile loading data, it incorporated six peak upper neck tensile loads the combined loading Cheng et al. frontal impact PMHS study (Cheng et al., 1982). With such a small sample size of PMHS ($N$ = 22), each data point has a significant influence on the curve. Both side impact and frontal impact often result in a kinematic response that includes neck flexion combined with tension, possibly explaining the similarity of the risk curves. From the FAA side impact study, the average tension for AIS 2+ injury was 2248N and for AIS 3+ injury was 2324N under accelerations ranging from 8.5-19 G. The average tension for AIS3+ injury from the ORG curve was 4247 N, with six of the ten subjects loaded at significantly higher accelerations ranging from 32-39 G. The PMHS data to construct the final (COM) curve for the present study had a mean tensile loading of 3523 N at the AIS3+ level. Thus, AIS 2+ injury generally starts at 2250 N and AIS 3+ injury generally between 2325 N and 4250 N. The equation for the COM risk function where T is neck tension in Newtons is:

$$P(AIS \geq 3) = \frac{1}{1 + e^{6.318 - 0.00297 * T}} \tag{10}$$

This study sought to develop and propose an updated USAF tensile neck injury criterion and compared it to the legacy criterion. The COM risk function for AIS 3+ injuries is recommended as the basis for an updated USAF tensile neck injury criterion, resulting in a 5%

injury protection limit for peak cervical spine tensile loading of 1136N. This injury criterion can be applied to the development of HMD's and assessing their safety for occupant use. This risk function is significantly more conservative than the criterion proposed by Carter et al. (Carter et al., 2000), which had a corresponding force of 5% injury tensile load of 1559 N. This difference is primarily due to the inclusion of recent PMHS cervical spine tensile failure data, which is substantially less censored. The updated risk function exhibits favorable characteristics to be used as the basis for a USAF tensile criterion. It incorporates the most up to date PMHS injury data available in the literature as well as the most comprehensive human subject testing tensile data and interprets this data using state of the art SA methods. The risk function produced in the present study is consistent with the tensile loading injury risk function used as a basis for the FAA's current neck injury criterion for side facing aircraft seats, providing verification on the adequacy of the results. The improved risk function, from which a USAF tensile neck injury criterion can be established at the desired risk level, is a significant contribution in the study of occupant protection.

# VII. Development of Neck Injury Criteria to Aid Development of Vehicle Safety Systems

## Chapter Overview

The paper that comprises this chapter will be submitted for publication to the Journal of Biomechanical Engineering in abbreviated form. It provides an overview of the methods to develop the three axis specific sub-elements (see **Figure 27**) that constitute the complete pilot scale multi-axial neck injury criteria (MANIC) developed in this research to aid design and test of HMD-centric escape systems. The performance of the MANIC application to real world escape system testing data is discussed and compared to the legacy criteria. Significant data gaps are identified, which would need to be addressed to move the MANIC from a pilot-scale set of criteria to a full-scale set of criteria for use as an USAF testing standard.

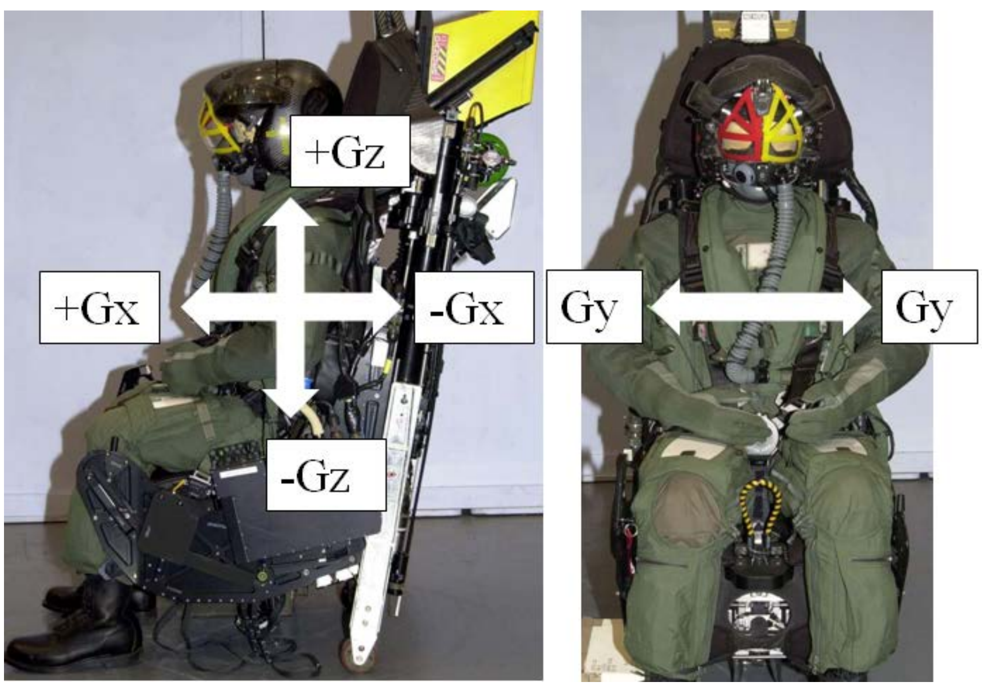


**Figure 27. The Three Primary Axes of Acceleration (Gx, Gy, and Gz)**

## Background

Injury risk functions with the capacity to predict the risk of a defined level of injury are important in the development of vehicle safety systems.  The risk function permits the designer to understand the relationship between observed loading and injury risk, enabling the designer to limit loading input to a level which corresponds to a maximum percent likelihood of a desired injury risk threshold.  In a general classification, minor injury typically involves only soft tissue damage, with no bone fractures (Bogduk and Yoganandan, 2001).  Major injury involves a fracture of the cervical spine or a neurologic injury which involves either the spinal cord or the nerve roots (Cusick and Yoganandan, 2002).  One standardized injury scale, the Abbreviated Injury Scale (AIS), is a clinical index of injuries that specifically defines the injury and assigns it a severity rating from 0 to 6 (AAM, 2008).  The AIS is commonly used in many injury criteria currently employed due to its exact delineation of the type of injury and its corresponding severity.  These features make it ideal for generating injury risk functions at specific AIS levels (e.g., AIS 3 or greater) for the purpose of limiting the injury.  While the AIS specifically classifies in detail and labels each injury with a severity, in general, AIS 1 is minor, AIS 2 is moderate, AIS 3 is serious, AIS 4 is severe, AIS 5 is critical, and AIS 6 is maximal.

Injury risk functions have been developed and applied for several vehicle domains, including automotive (Eppinger et al., 1999; Eppinger et al., 2000), civil aviation (FAA, 2011) and military aviation (Nichols, 2006).  Injury risk functions are valuable in the qualification of safety systems as they allow decision makers to design and evaluate systems to a specific level of acceptable risk (Pellettiere, 2012).  These functions are formed using various statistical techniques that model injury probability as a function of some input (Cutcliffe et al., 2012).  Further, these models define the risk of injury based upon analysis of experimental data with

either specific force input or a combination of forces for input and a pre-specified binary threshold outcome (injury/no injury) as the dependent variable.

While injury risk functions are useful in several application domains, the level of acceptable injury can differ between application domains. For example, in automotive applications, it is generally assumed that first responders will be on site shortly after a car accident to attend to any injury sustained in the collision and therefore the acceptable injury limits have been set to permit a 22% chance of an AIS 3 or lower injury (Eppinger et al., 1999). However, in military aviation, the safety system must protect against injuries that would result in incapacitation in a water ejection or significantly limit the pilot's ability to evade and escape if ejecting over a hostile area. Therefore, safety systems are often designed to avoid injuries above the AIS 2 classification level. In fact, guidance has been provided by the escape system acquisition oversight office of the Air Force Lifecycle Management Center (AFLCMC) that USAF aviation escape systems (also called ejection systems) should offer a level of protection that would result in no more than a 5% chance of an AIS 2 injury (Parr et al., 2013).

The development of improved criteria to evaluate the safety of aviation ejection-based escape systems has recently become important due to perceived limitations of existing methods as new ejection-based escape systems are being designed by the DoD. The renewed interest in improved criteria results from the development of systems where multiple subsystems are being simultaneously redesigned, additional head supported mass is potentially being added as helmet mounted displays (HMDs) are adopted, and the anthropometric diversity of the pilot population is increasing.

Helmet mounted displays (HMDs) are becoming common military human-machine interface equipment in manned flight. They have been shown to increase the performance of operators in their weapon systems and thus increase overall mission effectiveness by adding

capabilities such as enhanced night vision and information fusion. The benefits HMDs provide have the potential to enhance mission effectiveness across the spectrum of military operations (Rash et al., 2009). Unfortunately, this increased capability is often accompanied by increased mass, which can threaten pilot safety during ejection (Lewis, 2006; Nakamura, 2007; Stemper et al., 2009). Heavier HMDs worn for mission durations pose greater threat to the neck of pilots in an ejection than the lighter standard flight helmet. Injury risk due to a heavier HMD in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures, ligament ruptures, and spinal cord damage (Buhrman and Perry, 1994; Stemper et al., 2009).

Pilot anthropometric factors may also affect the likelihood of injury from neck loads induced by head supported mass. Smaller individuals and female pilots may be vulnerable to greater risk of injury. Recent changes in Department of Defense (DoD) pilot accommodation requirements have increased the range of pilot size from between 145 and 220 lb to between 103 and 245 lb (Harris, 1997). Therefore, it is important that pilot neck response be understood and characterized using a standard evaluation criteria that considers the influence of pilot anthropometric characteristics (e.g. mass and gender).

Injury risk posed by added head supported mass must also be understood when developing a new aircraft or a re-designed escape system for legacy aircraft. Acceptance testing is the step in the USAF acquisition process when complete escape system safety is measured and quantified using anthropomorphic test devices (ATD) to record neck loads and moments during multiple tests covering a range of ejection speeds (measured in knots equivalent airspeed or KEAS) and ATD mass (ranging from 103 lb to 245 lb). The purpose of this acceptance testing is to ensure acceptably safe ejection escape systems are fielded. The F-35 aircraft development program is the most recent instance of qualification testing for a newly developed escape system.

A recent example of qualification testing performed on an aircraft escape system modification program is the A-10 / F-16 aircraft helmet modification program, which added an HMD targeting system called the Helmet Mounted Integrated Targeting System (HMIT) to the existing flight helmet. The escape system oversight office of AFLCMC has promulgated requirements for AF aviation, specifying that neck injury criteria be developed to evaluate HMDs and new escape systems such that acceptable probability of injury be limited to 5% at an AIS 2 (moderate injury) (Parr et al., 2013). The 5% injury rate is a requirement for any single portion of the pilot population. For example, lower probabilities of injury observed in large males cannot be traded for higher probabilities of injury observed in small females. The 5% injury rate is also a requirement that should be met across the range of relevant airspeeds. The requirements also specify that multi-axial criteria be developed which are applicable for the full range of pilot size (103 to 245 lb). However, no existing neck injury criteria currently meet these requirements. This paper presents the development of multi-axial neck injury criteria to aid the design and evaluation of escape systems incorporating HMDs and applies these criteria to two sample escape system evaluation data sets. The criteria developed in this paper meet the AFLCMC neck injury criteria requirements.

Perhaps the best known criteria and accompanying neck injury risk curve has been developed by the National Highway Transportation Safety Administration (NHTSA) for frontal impact collisions (-Gx acceleration). This neck injury criterion, called the Nij, is applied in the United States as part of a comprehensive crash protection safety standard used in the assessment of advanced automotive restraint systems (Eppinger et al., 1999; Eppinger et al., 2000). The primary purpose of this criterion is to provide a consistent and quantitative method for evaluating and differentiating automotive crash and restraint systems where the quantitative metric (e.g., Nij) can be related to the likelihood of injury at a specified AIS level. This metric has a strong

foundation in biomechanics and relies upon results of crash tests with standardized ATDs to provide a criterion for predicting the likelihood of injury to persons with varying anthropometric characteristics for various automotive crash and restraint systems (Eppinger et al., 2000). The ability to define a relationship between the performance of the automotive crash and restraint system and the likelihood of injury, especially for persons with varying anthropometric characteristics, is a key attribute of the Nij criterion that is highly desirable.

The Nij established critical limits in four types of neck loading that are dominant in frontal impact automotive crashes; axial loading (tension and compression), and sagittal plane bending moments (flexion – forward, and extension – rearward). This criterion was developed using a methodology initially presented by Klinich et al. (Klinich et al., 1996). The researchers who developed this injury criterion applied previous biomechanical experiments using volunteer humans, porcine subjects and post mortem human subjects (PMHSs). This same research established critical limits for these four load pathways (Mertz et al., 1978; Nyquist et al., 1980; Mertz and Patrick, 1971; Yoganandan et al., 1996; Shea et al., 1992; Lenox et al., 1982) and methods to scale these critical limits to individuals with a broad range of body mass. The formula used to calculate the Nij is shown in Equation 11.

$$N_{ij} = \left|\frac{F_Z}{F_{Zcrit}}\right| + \left|\frac{M_Y}{M_{Ycrit}}\right| \qquad (11)$$

where

$F_Z$ = peak observed upper neck axial load (tension/compression)

$M_Y$ = peak observed upper neck sagittal plane (flexion/extension) bending moment

$F_{Zcrit}$ = axial load critical values (different for tension and compression)

$M_{Ycrit}$ = sagittal plane bending moment critical values (different for flexion/extension)

During qualification testing for a new automobile, neck loads *Fz* and *My* are observed at the upper neck (occipital condyles or OC) of an ATD during a standardized automotive crash scenario. The critical values were established by NHTSA for axial load in tension or compression and bending moment in flexion or extension (Eppinger et al., 1999). Different critical load values were established for each ATD representing individuals within different anthropometric categories (small female is 103 lbs, mid male is 172 lbs, and large male is 220 lbs). The "ij" subscript of the Nij signifies indices for the four combination mechanisms for injury, $N_{TE}$, $N_{TF}$, $N_{CE}$, and $N_{CF}$, where T and C represent the axial load index (tension or compression) and F and E represent the sagittal plane bending moment index (flexion or extension) (Eppinger et al., 1999). The current Nij "performance limit" is set at 1.0, meaning an automotive test that produces ATD neck loads that exceed an Nij value of 1.0 fails the criterion. An Nij of 1.0 represents a 22% risk of an AIS 3 or greater injury, considered a moderate injury (Eppinger et al., 1999). The risk curves associated with Nij values are an important part of the criterion as they provide likelihood of injury information and are covered in detail in Eppinger et al. (Eppinger et al., 1999). Unfortunately, the resulting injury risk curves were created using standard logistic regression and the AIS 2 curve intercepts the probability axis at a value greater than 5%. Therefore, these risk functions cannot be used to assess the 5% of an AIS 2 injury as required for military aviation (Parr et al., 2013). Additionally, the Nij provides a neck injury criterion for acceleration in the x-axis, while pilots can undergo high accelerations in any of the three cardinal axes during an ejection (Pellettiere et al., 2005). Therefore, multi-axial criteria are needed which capture the risk of injury as a result of y and z axis accelerative inputs.

A research team from the United States Naval Air Systems Command has put forth neck injury criteria (NIC) which include a family of metrics used to assess potential neck injuries in ejection, which will be referred to as the NIC (Nichols, 2006). The purpose of these criteria are

to evaluate the safety of the pilot during ejection using new escape systems to limit neck injury hazard to pilots to acceptable levels. The criteria are also applied to qualify new equipment introduced into an existing escape system. The NIC has most recently been used to evaluate new ejection seat acquisition programs (e.g. F-35), ejection seat modification programs (e.g. the T/AV-8B Ejection Seat Improvement Program and Naval Aircrew Common Ejection Seat Stability Improvement Program), and HMD programs (e.g. F-18A/B Joint Helmet Mounted Cueing System) (Nichols, 2006). The NIC incorporates 12 neck injury criteria which include six modes of neck loading evaluated at two locations in the neck, the upper neck (OC) and the lower neck (T-1/C-7 junction). The six modes of loading evaluated at both upper and lower neck in the NIC are 1) tension duration ($+F_Z$), 2) compression duration ($-F_Z$), 3) resultant shear duration ($F_{X,Y}$), 4) Nij (composite of simultaneous maximum tension/compression ($F_Z$) and peak flexion/extension ($My$)), 5) Neck Moment Index (NMI) for $Mx$ (maximum instantaneous lateral bending), and 6) NMI for $Mz$ (maximum instantaneous twisting). Each of these loads have associated limits. In general and where possible the NIC limits correspond to a 10% risk of AIS 3+ neck injury, but this correlation is unclear, both the probability of injury and the injury levels are not clearly supported by robust risk functions.

Unlike the Nij, the NIC considers the ejection neck injury criteria as “success criteria” rather than pass/fail criteria, due to the dynamic and complex nature of an ejection event (Nichols, 2006). The application of these criteria is described as a set of flags. If none of the criteria are failed during a test, then the test is a success with no caution flags raised. If one or more of the criteria are failed during a test, then a flag is raised and the issue is investigated by a panel of subject matter experts (SMEs) to determine if the failure indicates a potential cause of injury (Nichols, 2006). This is accomplished by reviewing the details of the exceedence including “body position, off axis neck loading, seat, chest, and head linear and angular

acceleration, the portion of the limit curve that was exceeded, and the magnitude of the exceedence (Nichols, 2006)." Depending on these details involved with an exceedence of one or more of the criteria in a test, the exceedence might be dismissed if it is considered low risk. Or it might be accepted if the details of occurrence support evidence that a neck injury hazard truly exists. The reader is referred to Nichols (2006) for further details.

The tension, compression, and shear force duration limits used in the NIC are based upon the Mertz automotive duration criteria, but have been modified for application to the ejection environment (Mertz, 1993). According to Nichols, the short duration tension limits correspond to about a 10% risk of AIS 3 neck injury, and while the longer duration load limits also correspond to some injury mechanism it is unspecified what this injury risk is in the NIC (Nichols, 2006). The risk of injury for the compressive duration limits and the shear duration limits are also not specified or known. The vague and indeterminate nature of the duration thresholds used in the NIC make their use, and more specifically their justification, difficult in the application of these neck injury criteria to acceptance testing, where timely and precise injury risk assessment based on observed neck loads and moments are crucial to acquisition decision makers.

The NIC is like the Nij in that it was designed to evaluate an ejection seat or component of the escape system based upon observed neck loads in an ATD. However, it only provides a prediction capability for the probability of various levels of AIS injury for one of the 12 criteria (upper neck Nij) for which risk curves have been developed. This risk curve provides decision makers the ability to choose an acceptable level of risk. The upper neck Nij portion of the criteria uses a risk curve developed by NHTSA, though these risk curves have been shown inadequate for the ejection environment (Eppinger et al., 1999; Parr et al., 2012). The other 11 sub-criteria, which have only load limits but no risk curves, only afford binary injury prediction

capability about seemingly arbitrary loads. The NIC is comprehensive in nature, incorporating multi-axial loading which is experienced by the pilot in the ejection environment that would potentially cause harmful loading to the neck, but the lack of explicit injury risk functions in 11 of the 12 sub-criteria is problematic. It the safety of various systems being developed like an ejection seat modification, addition of an HMD, or a completely new aircraft escape system to be evaluated against specific load thresholds, but does not provide useful risk information results , which could permit decision makers to consciously trade injury risk for other desired capabilities.

While the NIC sets comprehensive limits on potential pathways for injurious neck loading in the 12 elements of the criteria, some of these are redundant and conflicting. That is, it is possible for a system to pass the load duration tension or compression criteria of the NIC but fail the tension or compression criteria embedded in the Nij. This redundancy makes the criteria difficult to use for making tradeoff analyses during system design. It also makes it hard for program managers in the acquisition community to provide definitive requirements and specifications to their contractors. Other researchers have critiqued the NIC and have suggested changes to improve the criteria relative to the conflicting standards that make for difficult system evaluation (Carter et al., 2000; Pellettiere et al., 2011; Pellettiere, 2012).

Bass, Donnellan, Salzar and colleagues proposed a neck injury criterion called the Beam Criterion for the lower neck based upon accelerative testing of PMHSs with head supported mass in various frontal and vertical orientations (Bass et al., 2006). Their lower neck injury criterion is structured similarly to the Nij, based on a beam model of the lower cervical spine, though initially a shear component was included but later removed because it did not improve the predictive ability of the risk function. Based upon experimental evidence from 36 cadaveric head/neck complexes and six whole PMHSs, they observed that injury to the lower neck was

more prominent with the addition of head supported mass, and thus constructed their injury criterion based upon forces at the lower neck (Bass et al., 2006). Additionally, rather than a logistic regression, they applied survival analysis to develop risk curves based upon the fact that their data set consisted of censored data; injury tests were left censored and non-injury tests were right censored (Bass et al., 2006). The critical values used as a starting point in the beam criteria to scale the axial loads and sagittal plane bending moments were taken from the NHTSA Nij 50$^{th}$ percentile male Hybrid III ATD simple bending values (4170 N tension, 4000 N compression, and 190 N-m flexion) (Bass et al., 2006). Once a baseline risk function was produced, the researchers determined optimum critical values by allowing the ratio between the flexion and tension critical values to vary and constraining the mean 50% injury risk to equal 1.0 with standard deviation minimized (Bass et al., 2006). This resulted in the risk function statistically optimizing the critical values (new values of 5660 N tension, 5430 N compression, and 141 N-m flexion) (Bass et al., 2006).

Bass et al. compared the Nij evaluated at the upper neck with their criterion evaluated at the lower neck and concluded that based upon their experimental observations the Nij was not an adequate neck injury criterion in inertial loading with head supported mass (Bass et al., 2006). This is because they observed the overall kinematics of the Hybrid-III ATD to be significantly different from cadavers in the accelerative testing with head supported mass. The authors posit that since the Nij is built around the neck response recorded from the Hybrid-III, the resulting neck injury conclusions drawn from a Hybrid-III test with head supported mass are flawed (Bass et al., 2006). It was observed that the THOR ATD had kinematics that more similarly matched the cadavers in testing. However, the bulk of the data points involved in constructing the Beam Criterion are from PMHS segments potted such that they were only mobile from T2 and up (T3-T4 spinal segment was immobilized and potted into a mounting fixture). This may have caused

the kinematic response to be different from a whole PMHS, potentially affecting the results. Salzar et al. found the Beam Criterion to be less accurate in predicting injury in small PMHS accelerative +Gz sled tests (Salzar et al., 2009) compared to the Nij and the NIC.

Criteria have also been developed which apply methods similar to those applied during derivation of the Nij to establish criteria for side facing aircraft seats (i.e., acceleration input in the y axis) (FAA, 2011), as well as a tensile loading criterion in the z-axis (Carter et al., 2000). While each of these criteria is linked to injury risk curves, they each included relatively small sample sizes and logistic regression in deriving the injury risk function.

Logistic regression (LR) is commonly used in data analysis where researchers desire to model an association between a binary or dichotomous response variable and one or more predictor variable(s) (Hosmer and Lemeshow, 2000) and has been used in the literature to generate injury risk functions (Eppinger et al., 1999; FAA, 2011; Carter et al., 2000). However, this approach assumes a large sample size ($N \geq 100$), and that each data point is exact. In the field of injury biomechanics, studies commonly involve small ($N \leq 100$) samples of human subjects and PMHS, and often the sub-injurious sample size is far greater than the injurious sample size. Also, exact data may not exist as observations may be taken such that the conditions at the time of injury is not known (just that injury did occur), or that testing is stopped early such that injury could not occur. A method which can produce appropriate regression estimates using smaller data sets is Firth's adjusted maximum likelihood method. This method may be used as a correction where LR coefficients might be biased when data is skewed toward one outcome (injury or no injury). In addition, this method is useful when there is a small sample size ($N \leq 100$), or the contingency table (for discrete predictors and outcomes) has too many cells with low counts (Firth, 1993). Although, this method may be used to make the LR

model more appropriate for such data, methods designed specifically for this type of data would be optimal.

An alternative statistical method for risk function construction is survival analysis (Hosmer et al., 2008), which is becoming a prevalent method to generate injury risk functions in the field of injury biomechanics (Cutcliffe et al., 2012; Bass et al., 2006; Parr et al., 2013). The increased use of this method is partially due to advances in computing capability and the incorporation of survival analysis techniques in “point and click” statistical software and the ability of survival analysis to appropriately handle censored data. In the field of biomechanics, data is often gathered in such a way that the exact value of an observed neck loading is unknown. In the case of injurious testing, the actual value of the loading that caused the injury in the PMHS may be less than the loading value recorded (Cutcliffe et al., 2012). This type of data is referred to as being left-censored. Alternatively, for non-injurious human subject testing, the actual value of the loading that might cause injury is greater than the loading value recorded. This type of data is referred to as being right-censored. An injury risk function developed with data from both human and PMHS data would be using both left and right censored data. SA may be appropriately used with censored data in order to produce injury risk functions.

Although a set of robust neck injury criteria are required for military aviation ejection systems, the extant criteria do not provide multi-axial criteria with supported neck injury risk curves capable of assessing a 5% risk of an AIS 2 injury. The current research relies upon existing data from previously conducted experiments with human participants and PMHS to construct multi-axial criteria with neck injury risk curves. While the scope of the present work is limited by available data, it is intended that the approach will identify the areas where additional data collection should be undertaken as well as provide a basis for a robust criteria as additional data becomes available.

## Methods

### Criteria Formulation

In the current study, risk functions were constructed with combined human subject and whole specimen PMHS data for each axis of lab testing acceleration applicable to an analogous component of the complete ejection sequence where adequate data was available. The human subject data provides sub-injurious neck load data and comes from previous experiments performed at Wright Patterson Air Force Base at the Air Force Research Laboratory (AFRL) Biomechanics Branch. The specific human subject data used for the –Gx, Gy, and –Gz risk functions were selected because they represented the highest loading conditions available (from both head supported mass and accelerative input) in the AFRL biodynamics data base. The PMHS data provides the injurious neck load and injury data and was sourced from the literature or through partnerships with other research institutions.

For the present risk function construction it will be assumed that instantaneous loads observed in the upper neck can be applied to establish robust neck injury functions. For data to be considered adequate in this case it must satisfy a few important requirements. First, for each accelerative event the subject's peak upper neck (OC) loading must be provided, either via raw time history load data or published peak observed load data. Second, specific injury levels classified using the AIS scale that resulted from the loading are required for PMHS experiments. Although in application, inputs may occur in each primary axis or a combination, to simplify data collection and formulation, separate criteria will be derived for each of the axes, assuming independence. It will be assumed that the risk present in any system the criteria are applied to is the maximum risk present as a result of an accelerative input in of the three axes.

Data availability drove the resulting makeup of the combined load equation for each axis of acceleration. It is assumed that any of the three inputs may result in a neck force or a moment

about the occipital condyles. Therefore, it will be assumed that the final criteria should include all six of these elements if data is available to support the inclusion of all six. However, in the interim metric, it may be necessary to adopt fewer forces and moments when data is lacking to support the development of a model including all six resulting forces or moments. To construct human-based risk functions, the data sets for each axis were required to include injury and non injury data points from experiments with similar conditions.

To summarize the available data, adequate human subject and PMHS data were available to construct risk functions incorporating one or more neck loads for -Gx (frontal impact acceleration), Gy (side impact acceleration), and -Gz (tensile loading acceleration) experiments. No known PMHS or human subject, ejection-like +Gx (rear impact acceleration) data exist in the literature. The +Gx axis of acceleration is not a commonly experienced mode in ejection because escape systems are designed to maintain a front facing orientation throughout the ejection sequence. However, a significant gap that exists in the literature preventing the construction of multi-axial neck injury criteria is PMHS +Gz (vertical seat accelerative impact) neck data. This would be analogous to the catapult phase of ejection and is a significant concern of escape system engineers and users (Salzar et al., 2009). To the authors' knowledge no experimental PMHS data exists in the literature in the +Gz axis of acceleration that contains some or all of the six primary neck loads (*Fx*, *Fy*, *Fz*, *Mx*, *My*, *Mz* time history data) and corresponding AIS injury classification. A very small sample size (N=3) +Gz experiment has been performed with 5$^{th}$ percentile matched pair PMHS and ATD (Salzar et al., 2009). However, actual PMHS neck loads were not estimated, only the corresponding matched ATD neck loads were reported.

Neck injury due to ejection seat catapult (+Gz acceleration) has historically been a less significant safety concern until the recent addition of HMDs to the pilot ensemble (Lewis, 2006).

Lumbar and thoracic spinal injuries have been the main concern, and criteria to protect against spinal injury in this region have been adequately developed and implemented (NATO, 2007). Adequate human subject data is available in the +Gz axis of acceleration, though only four of the six primary loads are available (*Fx*, *Fy*, *Fz*, and *My*). The other two loads (*Mx* and *Mz*) were not observed at the time of the tests due to the limited space available for accelerometers on the bite bar used to record head accelerations, which were then translated into forces and moments based upon subject anthropometry. Additionally, *Mx* and *Mz* were not observed to be significant in this mode of acceleration, thus the investigators decided not to record them.

Other fundamental elements of the structure and nature of the criteria include the following. For the criteria developed an upper neck, peak loading criteria was selected rather than a load duration criteria. This is based upon the observation that upper neck, peak loading criteria has been established in the literature as the current state of the art preferred type of risk criteria in the injury biomechanics field (FAA, 2011; Eppinger et al., 2000). Additionally, there is no established method in the literature to experimentally develop risk functions at quantified injury levels for a duration loading criteria. It was also determined that SA is the most appropriate statistical method to develop the risk function based upon the use of censored load-to-failure data. While the USAF escape system oversight office desires risk criteria which limit injury risk at the AIS 2+ level, both AIS2+ and 3+ risk functions were constructed to demonstrate the robust ability of this method to quantify risk at various levels of injury. For each input axis, it will be assumed that the forces and moments experienced by the neck are independent of one another and that the risk of injury is not affected by any interactions among any of these forces or moments. This study adopted the use of critical values used in the Nij (Eppinger et al., 2000) which have been adapted and used in the ejection environment to normalize the forces and moments incorporated into the combined loading formulations

(Nichols, 2006). As it makes sense, a root sum of squares formulation will be adopted as proposed by Perry and colleagues (Perry et al., 1997) because it removes negative values and allows for the total response to be dominated by larger values in the input variables. This latter attribute of the root sum of squares formulation is important as it serves the desired purpose to capture the important neck load responses in each axis of acceleration. However, due to the prominence of the Nij formulation, the sum of absolute values formulation will be adopted in the Gx axis if the final formulation includes the same loads as the Nij.

The ideal MANIC formulation is shown in Equation 12. At the start of the current study it was desired that a complete 6-load multi-axial structure be used for the independent variable of the risk function. This structure includes all six primary neck loads in a root sum of squares formulation, called the multi-axial neck injury criteria (MANIC) after Perry et al. (Perry et al., 1997). The denominators for each component force that comprise the MANIC are critical values that scale and normalize each force based upon known component neck strength and occupant size. For example, the neck is stronger in flexion (+*My*) than extension (-*My*) and thus the critical value is higher for flexion than it is for extension (Eppinger et al., 1999). In a combined loading neck injury criterion structure such as the MANIC, the critical values are very important to ensure that the proper weight is assigned to each load and the individual contribution of each load relative to the other loads is appropriately reflected in the combined output. As these critical values are scaled with occupant mass, they also generally normalize for differences between individuals. This scaling is also important to ensure that the contribution of each load is appropriate for each subject's mass.

$$MANIC = \sqrt{\left(\frac{Fx}{Fxcrit}\right)^2 + \left(\frac{Fy}{Fycrit}\right)^2 + \left(\frac{Fz}{Fzcrit}\right)^2 + \left(\frac{Mx}{Mxcrit}\right)^2 + \left(\frac{My}{Mycrit}\right)^2 + \left(\frac{Mz}{Mzcrit}\right)^2} \quad (12)$$

where

$Fx$ = observed x direction shear loading

$Fxcrit$ = critical intercept value for x direction shear loading

$Fy$ = observed y direction shear loading

$Fycrit$ = critical intercept value for y direction shear loading

$Fz$ = observed axial loading ($+Fz$ = tension, $-Fz$ = compression)

$Fzcrit$ = critical intercept value for axial loading (different for tension/compression)

$Mx$ = observed moment about the anatomical x axis (side bending)

$Mxcrit$ = critical intercept value for side bending

$My$ = observed moment about the anatomical y axis (sagittal plane anterior/posterior bending, $+My$ = flexion, $-My$ = extension)

$Mycrit$ = critical intercept value for sagittal plane moments (different for flexion/extension)

$Mz$ = observed moment about the anatomical z axis (neck twisting)

$Mzcrit$ = critical intercept value for neck twisting

**Figure 28** illustrates the general steps taken in this research to move from laboratory accelerative experiments in each primary axis with human and PMHSs to a set of criteria limits that can be applied to observed neck loads from a system test to assess system safety performance. As shown in **Figure 28**, a human subject or PMHS is exposed to input acceleration in a single axis, which causes the human body to flex such that the head undergoes

acceleration in the three linear axes as well as the three rotational axes, resulting in forces and moments about the occipital condyles. These forces and moments are used to determine MANIC values at each time step in the experiment and the peak MANIC value is determined for the experiment. This MANIC value and the injury classification are used (along with data points from other tests) as the input data to create an axis-specific injury risk function. From the risk functions, the user can define injury risk limits based upon the desired percentage of injury risk allowable for the specific application. The risk function provides the max allowable MANIC(Gx), MANIC(Gy), and MANIC(Gz) values based upon the acceptable injury risk determined by the user. The three-axis set of limits stating the maximum allowable load in each axis taken together constitutes the complete MANIC.

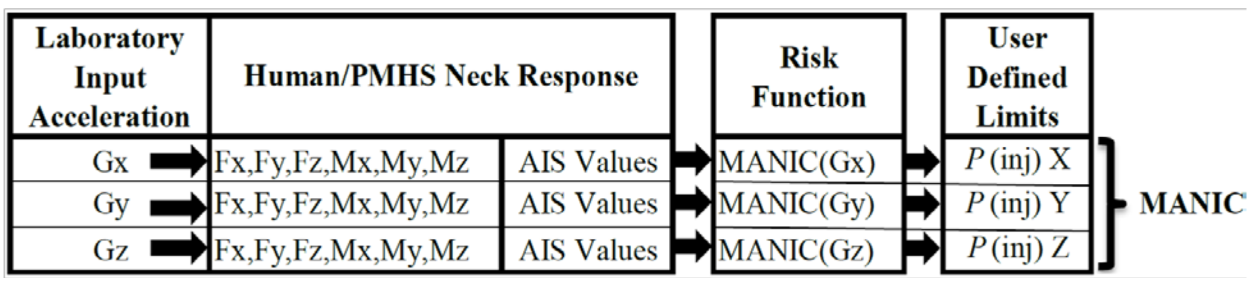


**Figure 28. MANIC Development Process**

**MANIC(-Gx) Risk Function Construction**

Data from a previously performed human subject experiment on the effects of variable helmet mass on neck response to -Gx acceleration (Doczy et al., 2004), which might represent the acceleration sustained from a frontal automotive impact or parachute opening phase of ejection, was used as the non-injurious portion of the data set. The test “HMD” was a standard USAF flight helmet (HGU-55/P) modified to allow variable mass to be attached to the helmet, which was properly fitted and attached to the subject’s head using standard chin straps. For ease of reference this test helmet will subsequently be referred to as the HMD.

Data from three AFRL human subject experimental test configurations were used in this analysis. These experiments contained conditions which resulted in the highest non-injurious neck loading available without exposing the participants to more than minimal risk. In the first test configuration, 26 human subjects wearing a 2 kg HMD were subjected to 6 Gs of acceleration. In the second test configuration, 24 subjects wore a 1.6 kg HMD and were subjected to 8 Gs of acceleration, and in the third test configuration, 23 subjects wore a 2 kg HMD and were exposed to 8 Gs of acceleration.

During the test, volunteer subjects were seated vertically and restrained in a standard USAF fighter aircraft ACES-II ejection seat. The seat was mounted to the test sled and subjects were accelerated rearward on the sled track at the specified acceleration level to measure the -Gx neck responses. A tri-axial linear accelerometer and an angular accelerometer mounted on a bite bar measured the head accelerations (Doczy et al., 2004). The accelerative portion of the experiment lasted for about 200 ms. All of the tests were non-injurious but neck stiffness or soreness (classified as less than AIS 1 injuries) was reported in approximately 15% of the tests, mostly at the higher helmet mass and acceleration levels (Doczy et al., 2004). Human subject neck loads were computed using subject anthropometry, exact helmet inertial properties, and bite bar-recorded head accelerations at ms increments (Doczy et al., 2004; Gallagher et al., 2007).

The peak upper neck load data from these three human subject test conditions (N=67) were combined with the largest known set of adequately characterized injurious PMHS data (N=6) (Cheng et al., 1982) to generate the MANIC(Gx) AIS 2+ and 3+ risk functions using a parametric survival analysis assuming a logistic distribution following established methods (Parr et al., 2013; Bass et al., 2006). Detailed information for the human subjects in each of the tests is shown in **Table 17**.

The upper neck (OC) load data available from these human subject tests were *Fx*, *Fy*, *Fz*, and *My*. The upper neck (OC) load data available from the Cheng et al. PMHS study was *Fz* and *My* (Cheng et al., 1982). Thus the loads available from both data sets from which to construct the MANIC(Gx) risk function were *Fz* and *My*. Based upon the available data the structure used to calculate the normalized peak combined loading for each individual human subject and PMHS experiment is a combination of *Fz* and *My* shown in Equation 13 (the same as the Nij formulation).

$$MANIC(Gx) = \left|\frac{F_Z}{F_{Zcrit}}\right| + \left|\frac{M_Y}{M_{Ycrit}}\right| \tag{13}$$

The absolute value is incorporated because +*Fz* is tensions and –*Fz* is compression and each has a different critical value. The same is true for *My*, where +*My* is flexion and –*My* is extension and each has a different critical value. The pure sum formulation has a robust biomechanical history in use in the automotive regulatory community in NHTSA's Nij (Eppinger et al., 2000) and since data is not currently available to add additional terms to this equation this form of the equation was maintained to permit comparison to the prior literature.

**Table 17. MANIC(Gx) Human Subject Anthropometry and Peak Instantaneous Upper Neck Loads**

| | | | | | | | Test Conditions | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Subject Anthropometry | | | | | | 8 G, 2 kg HMD | | | 8 G, 1.6 kg HMD | | | 6 G, 2 kg HMD | | |
| | Body Mass (kg) | Gender | Ht (cm) | Sit Ht (cm) | Age | Neck Circ (cm) | My (N-m) | Fz (N) | MANIC (Gx) | My (N-m) | Fz (N) | Nij | My (N-m) | Fz (N) | MANIC (Gx) |
| | 54.9 | F | 154.9 | 81.9 | 20 | 32.5 | - | - | - | 34.4 | 182.7 | 0.26 | 24.8 | 2.7 | 0.16 |
| | 60.8 | F | 158.8 | 84.3 | 29 | 31.2 | 46.6 | 152.7 | 0.17 | 24.8 | 153.0 | 0.10 | 15.4 | 115.2 | 0.07 |
| | 65.3 | F | 160.0 | 88.3 | 28 | 32.9 | 44.6 | 192.0 | 0.17 | 43.6 | 321.6 | 0.19 | 23.9 | 61.5 | 0.09 |
| | 65.8 | F | 175.3 | 91.4 | 19 | 31.5 | 32.8 | 230.3 | 0.14 | 32.6 | 203.7 | 0.14 | 48.4 | 41.9 | 0.16 |
| | 66.2 | M | 172.7 | 91.4 | 24 | 36.9 | - | - | - | 27.1 | 240.1 | 0.12 | 19.5 | 45.7 | 0.07 |
| | 68.0 | F | 165.1 | 84.5 | 27 | 33.0 | - | - | - | 40.6 | 15.4 | 0.13 | 25.2 | 188.4 | 0.11 |
| | 69.9 | F | 165.1 | 88.3 | 46 | 31.9 | 32.9 | 126.3 | 0.12 | 39.3 | 470.7 | 0.20 | 28.0 | 32.0 | 0.10 |
| | 72.6 | F | 170.2 | 88.9 | 23 | 35.8 | 40.5 | 111.2 | 0.15 | 34.7 | 122.8 | 0.13 | 24.8 | 12.6 | 0.08 |
| | 73.5 | F | 167.6 | 87.6 | 28 | 33.1 | 39.5 | 379.1 | 0.18 | 31.2 | 183.6 | 0.13 | - | - | - |
| | 73.5 | M | 180.3 | 94.0 | 35 | 36.7 | - | - | - | - | - | - | 18.1 | 3.0 | 0.06 |
| | 73.9 | F | 175.3 | 95.3 | 25 | 32.2 | 19.9 | 103.5 | 0.08 | 23.5 | 103.0 | 0.09 | - | - | - |
| | 73.9 | M | 188.0 | 100.3 | 30 | 35.5 | 30.0 | 12.3 | 0.10 | - | - | - | 22.9 | 3.3 | 0.07 |
| | 77.1 | M | 177.8 | 88.9 | 24 | 35.2 | 33.4 | 221.8 | 0.14 | 28.6 | 7.9 | 0.19 | - | - | - |
| | 78.5 | M | 177.8 | 96.5 | 27 | 38.3 | 30.3 | 2.3 | 0.10 | 32.7 | 1.5 | 0.11 | - | - | - |
| | 78.5 | M | 180.3 | 95.3 | 36 | 38.1 | 40.3 | 170.4 | 0.16 | 28.8 | 77.6 | 0.10 | 27.8 | 13.0 | 0.09 |
| | 81.6 | M | 175.3 | 87.6 | 30 | 37.9 | 33.8 | 141.1 | 0.13 | 28.2 | 333.4 | 0.14 | 25.2 | 38.2 | 0.09 |
| | 81.6 | F | 157.5 | 87.0 | 23 | 36.8 | 38.4 | 18.4 | 0.13 | 52.6 | 202.1 | 0.20 | 29.2 | 13.8 | 0.10 |
| | 83.0 | F | 172.7 | 90.2 | 29 | 35.9 | 41.8 | 89.4 | 0.15 | 25.8 | 6.1 | 0.08 | 22.5 | 4.8 | 0.07 |
| | 83.0 | M | 185.4 | 97.2 | 28 | 38.2 | 35.8 | 430.6 | 0.18 | - | - | - | - | - | - |
| | 84.8 | M | 180.3 | 94.0 | 31 | 39.2 | 28.9 | 393.0 | 0.15 | 35.2 | 238.3 | 0.15 | - | - | - |
| | 88.5 | M | 182.9 | 96.5 | 27 | 38.8 | 31.5 | 129.5 | 0.12 | 39.6 | 4.5 | 0.13 | 23.3 | 11.4 | 0.08 |
| | 89.8 | M | 185.4 | 95.3 | 22 | 36.8 | - | - | - | - | - | - | 36.2 | 23.9 | 0.12 |
| | 90.7 | M | 181.6 | 96.5 | 37 | 39.7 | 37.0 | 429.8 | 0.18 | 34.0 | 3.2 | 0.11 | 26.6 | 88.3 | 0.10 |
| | 90.7 | M | 182.9 | 96.5 | 36 | 39.8 | 35.5 | 3.8 | 0.09 | 39.9 | 9.1 | 0.10 | - | - | - |
| | 99.8 | M | 189.2 | 99.1 | 33 | 39.8 | 48.8 | 228.8 | 0.15 | 28.5 | 449.4 | 0.12 | - | - | - |
| | 119.7 | M | 185.4 | 97.8 | 36 | 45.3 | 42.3 | 0.1 | 0.10 | 39.4 | 1.6 | 0.10 | - | - | - |
| | 126.1 | M | 193.0 | 97.8 | 32 | 45.0 | 64.6 | 10.1 | 0.16 | 41.8 | 72.4 | 0.11 | 28.1 | 7.1 | 0.07 |
| **Mean** | 80.4 | N/A | 175.6 | 92.3 | 29.1 | 36.6 | 37.7 | 162.6 | 0.14 | 34.2 | 148.0 | 0.14 | 26.1 | 39.3 | 0.09 |

Empty cells signify that the subject did not participate in designated experimental condition

The six whole specimen PMHS data points were taken from previous research published by Cheng et al. (Cheng et al., 1982). This data set provides the largest published, whole specimen, frontal impact research available which included both observed neck loads and injury level. Frontal impact acceleration levels in this experiment were very high, between 32 and 39 G. Peak observed neck loads were estimated using acceleration and head mass to calculate forces. Injury caused by the impact was determined by autopsy and specified on the AIS scale.

Of the six PMHS, four experienced injuries classified as AIS 2 or greater (AIS2+), and three experienced injuries classified as AIS 3 or greater (AIS 3+) (Cheng et al., 1982). Thus any pair of risk curves generated for AIS 2+ injury and for AIS 3+ injury differs by a single injurious data point.

**Table 18. MANIC(Gx) Post Mortem Human Subject Load and Anthropometry Data**

| MANIC(Gx) | Neck AIS | Age | Max Sled G | Gender | Mass (kg) |
|---|---|---|---|---|---|
| **0.60** | 0 | 56 | 38.0 | M | 96.0 |
| **1.22** | 2 | 63 | 36.0 | M | 72.5 |
| **1.28** | 6 | 68 | 37.5 | M | 93.0 |
| **1.28** | 6 | 66 | 32.0 | M | 72.5 |
| **1.98** | 0 | 67 | 39.0 | M | 60.0 |
| **3.80** | 6 | 54 | 37.0 | F | 50.0 |

The *Fz* and *My* values used for the regression for the human subjects were the peak instantaneous value of the combined axial and bending loads. Unfortunately no time history was published for the PMHS data. Thus, only the peak individual values were reported and applied for axial loads and bending moments. Note that these forces did not necessarily occur at the same time. Because of this, the independent peak injurious PMHS MANIC(Gx) values are potentially higher than the peak instantaneous values would have been had they been observed. Thus, the resultant risk function is potentially slightly biased towards higher MANIC(Gx) values. The individual subject MANIC(Gx) values were calculated using the published NHTSA $N_{ij}$ intercept values (Eppinger et al., 2000) based upon occupant size by applying the small sized female intercept for subjects with body mass less than 63.5 kg the mid-sized male intercept values for subjects with body mass between 63.5 kg and 90 kg and the large male intercept values for subjects with body mass greater than 90 kg. Risk functions were generated through

parametric survival analysis (Hosmer et al., 2008) following the methods used in research by Bass et al. (Bass et al., 2006).

### MANIC(Gy) Risk Function Construction

Risk functions were constructed using a combination of Gy human subject data paired with Gy PMHS data. Two human subject data sets from a previous Gy acceleration experiment were used in this study (Perry et al., 2003). They were chosen because they provided the highest lateral neck load exposure of the experiments that have been performed to date at the AFRL laboratory accelerator test sled. Subjects were restrained in an ejection seat representative of operational USAF aircraft and subjected to a lateral (Gy), half-sine accelerative pulse with rise time and pulse duration of 75 and 150 ms respectively. The first study subjected 31 participants (21 male, 10 female) to 6 Gs of lateral acceleration (~5.5 m/s) with 1.36kg (3lb) of head supported mass. The second study subjected 25 subjects (17 male, 8 female) to 5 Gs of lateral acceleration (~4.6 m/s) with 2kg (4.5lb) of head supported mass. The typical kinematic response of the human subjects to Gy acceleration observed in slow motion video footage was an initial combination of neck twisting moment (*Mz*) and coronal moment (side bending or *Mx*) with the addition of flexion (*+My*) to this combination near the end of the accelerative pulse. Pure coronal moment (side bending) was not observed as a result of Gy acceleration.

The PMHS data use to construct the Gy risk functions were from research that supported the development of a neck injury criterion for side facing aircraft seats by a team of researchers from the Medical College of Wisconsin, Wayne State, and the FAA (FAA, 2011). From this data set, time history upper neck load data was available from 9 PMHS experiments subjected to Gy acceleration that ranged from 8.5-19 G. The subjects were placed into one of three different

test seating configurations representative of typical side-facing aircraft seats and restraints (FAA, 2011). Upper neck loads were calculated based upon observed head acceleration and subject anthropometry. Injury assessment post-test was categorized using AIS values, and ranged from AIS levels 0 to 5 (AAM, 2008). The PMHS experiment data table is provided in **Table 19**. For additional detail on the test set up, screening procedures and PMHS anthropometry the reader is referred to the final FAA summary report (FAA, 2011).

**Table 19. MANIC(Gy) Post Mortem Human Subject Load and Anthropometry Data**

| Subject | Mass (lb) | Crit Value (lb) | Peak MANIC(Gy) | Acceleration (G) | AIS |
|---|---|---|---|---|---|
| PMHS 1 | 138.8 | 136 | 0.85 | 15.5 | 2 |
| PMHS 2 | 142.0 | 136 | 1.99 | 12.5 | 5 |
| PMHS 3 | 147.7 | 150 | 0.63 | 15.5 | 5 |
| PMHS 4 | 154.0 | 150 | 0.41 | 12.5 | 1 |
| PMHS 5 | 163.0 | 172 | 0.72 | 19.0 | 1 |
| PMHS 6 | 164.0 | 172 | 0.27 | 8.5 | 0 |
| PMHS 7 | 167.0 | 172 | 1.60 | 12.5 | 5 |
| PMHS 8 | 180.0 | 172 | 0.27 | 8.5 | 3 |
| PMHS 9 | 190.0 | 200 | 0.35 | 12.5 | 1 |

Data availability necessitated a modified structure from Equation 12. The human subject experiment from which the data was collected was performed before data collection technology (e.g. bite bar sensors) were small enough to accommodate accelerometers to observe all six primary OC neck loads (*Fx*, *Fy*, *Fz*, *Mx*, *My*, and *Mz*). Due to the lack of observed *Mx* in human subject kinematics by the researchers at the time of the original human subject experiment it was decided that angular acceleration about the x-axis would not be recorded. Thus *Mx* (side bending) data was not recorded. As a result, a modified formulation with five of the six primary

neck loads (*Mx* excluded) was used to compute the peak instantaneous MANIC(Gy) as seen in Equation 14, referred to as MANIC(Gy).

$$MANIC(Gy) = \sqrt{\left(\frac{Fx}{Fxcrit}\right)^2 + \left(\frac{Fy}{Fycrit}\right)^2 + \left(\frac{Fz}{Fzcrit}\right)^2 + \left(\frac{My}{Mycrit}\right)^2 + \left(\frac{Mz}{Mzcrit}\right)^2} \qquad (14)$$

The critical values used in this axis are values that have been used in a recent DoD escape system qualification testing program (Nichols, 2006). These values incorporate data from the NHTSA Nij neck injury criteria (Eppinger et al., 2000) as well as Navy escape system qualification testing neck injury criteria (Nichols, 2006) and are scaled for ATD mass. Based upon a lack of alternative critical values in the literature, the ATD critical values were applied to human subjects as described in **Table 20** as a first order approximation (for example, 150 lb intercept values would be used for a human subject with a mass from 143 lbs to 161 lbs). The Nij has established critical values for +/- *Fz* and +/- *My* and are incorporated in **Table 20** but scaled for eight categories of individual body mass rather than the three categories used by NHTSA's Nij. For the forces that are not included in the Nij (*Fx, Fy, Mx, and Mz*), the critical values are based upon preliminary estimates of appropriate thresholds determined to limit injury in the ejection environment (Nichols, 2006). **Table 20** shows the intercepts used to calculate the MANIC(Gy) based upon subject body mass.

## Table 20. MANIC(Gy) Upper Neck Critical Values Based Upon Body Mass

| | | | Force | | | Moment | |
|---|---|---|---|---|---|---|---|
| **ATD Mass (lbs)** | **Human Mass (lbs)** | **Component** | **lb** | **N** | **Component** | **in-lb** | **N-m** |
| | | $F_{xcrit}/F_{ycrit}$ | 405 | 1802 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 593 | 67 |
| 103 | <114 | $-F_{zcrit}$ (Comp) | 872 | 3880 | $+M_{ycrit}$ (flexion) | 1372 | 155 |
| | | $+F_{zcrit}$ (Tens) | 964 | 4287 | | | |
| | | $F_{xcrit}/F_{ycrit}$ | 496 | 2206 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 845 | 95 |
| 125 | 114-130.5 | $-F_{zcrit}$ (Comp) | 1099 | 4889 | $+M_{ycrit}$ (flexion) | 1939 | 219 |
| | | $+F_{zcrit}$ (Tens) | 1214 | 5400 | | | |
| | | $F_{xcrit}/F_{ycrit}$ | 522 | 2322 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 912 | 103 |
| 136 | 130.5-143 | $-F_{zcrit}$ (Comp) | 1157 | 5147 | $+M_{ycrit}$ (flexion) | 2094 | 237 |
| | | $+F_{zcrit}$ (Tens) | 1278 | 5685 | | | 0 |
| | | $F_{xcrit}/F_{ycrit}$ | 561 | 2495 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1016 | 115 |
| 150 | 143-161 | $-F_{zcrit}$ (Comp) | 1243 | 5529 | $+M_{ycrit}$ (flexion) | 2333 | 264 |
| | | $+F_{zcrit}$ (Tens) | 1373 | 6107 | | | 0 |
| | | $F_{xcrit}/F_{ycrit}$ | 625 | 2780 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1195 | 135 |
| 172 | 161-186 | $-F_{zcrit}$ (Comp) | 1385 | 6160 | $+M_{ycrit}$ (flexion) | 2744 | 310 |
| | | $+F_{zcrit}$ (Tens) | 1530 | 6806 | | | |
| | | $F_{xcrit}/F_{ycrit}$ | 683 | 3038 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1364 | 154 |
| 200 | 186-210 | $-F_{zcrit}$ (Comp) | 1513 | 6730 | $+M_{ycrit}$ (flexion) | 3133 | 354 |
| | | $+F_{zcrit}$ (Tens) | 1671 | 7433 | | | |
| | | $F_{xcrit}/F_{ycrit}$ | 777 | 3456 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1584 | 179 |
| 220 | 210-232.5 | $-F_{zcrit}$ (Comp) | 1673 | 7440 | $+M_{ycrit}$ (flexion) | 3673 | 415 |
| | | $+F_{zcrit}$ (Tens) | 1847 | 8216 | | | |
| | | $F_{xcrit}/F_{ycrit}$ | 836 | 3719 | $M_{xcrit}/-M_{ycrit}$ (extens)/$M_{zcrit}$ | 1850 | 209 |
| 245 | 232.5+ | $-F_{zcrit}$ (Comp) | 1853 | 8243 | $+M_{ycrit}$ (flexion) | 4248 | 480 |
| | | $+F_{zcrit}$ (Tens) | 2047 | 9106 | | | |

Risk functions were constructed to predict AIS 2+ and AIS 3+ injury using a parametric survival analysis assuming a logistic distribution following established methods (Parr et al., 2013; Bass et al., 2006). The time history data of each subject's accelerative test was processed. The AFRL data provided the sub-injurious data points for the risk function, while the FAA side impact PMHS data provided the injurious data points. First, the unitless MANIC(Gy) was computed at each step in the time history of each subject's test run. Only neck load data observed prior to the head striking the head rest were used. Any neck load values recorded after a head strike would be inaccurate due to the effect head impact has on the measured head accelerations used to calculate the neck loads and thus were not used. The appropriate intercepts based upon the subjects' body mass from **Table 20** were applied in the computation of the MANIC(Gy). Then, the peak MANIC(Gy) value and the corresponding level of injury observed during the test according to the AIS scale were used to generate a data set consisting of peak MANIC(Gy) values and injury assessment. In this case, injury risk functions built to evaluate both AIS 2+ and 3+ injuries were desired. Thus the injury assessment was binary; either the subject did or did not experience an AIS 2+ or 3+ injury. The survival analysis was performed on the combined human subject and PMHS data set to construct an AIS 2+ and an AIS 3+ risk function.

### MANIC(-Gz) Risk Function Construction

For the Gz axis of acceleration, tensile load (+*Fz*) was the only common neck load between available human subject and PMHS data, thus a tensile only neck injury risk function resulted, which will be referred to as MANIC(Gz). The human subject experimental neck load data used to construct the risk function came from previous AFRL tests where the subject was

seated in a test ejection seat, but oriented horizontally (Brinkley and Getschow, 1988). Acceleration was applied to the seat such that the subject's body was accelerated away from their head, resulting in a neck response that was observed to be primarily tensile loading of the cervical spine. Tension values for the 208 human subject experiments were between 34 and 149 N with a mean of 77.2 N and a standard deviation of 18.3 N. No new data from human subject experiments in this orientation have been published since, therefore this data remains the best source of human subject neck tension data.

The PMHS tensile loads and injury classifications used to construct the MANIC(Gz) risk functions came from multiple sources to constitute an adequate sample size. Four separate studies consisting of 12, six, three, and one subject for a total of 22 PMHS tensile data points were used. Of the 22, two were non-injurious and 20 were injurious at an AIS level of 2 or greater; three were non-injurious and 19 were injurious at an AIS level of 3 or greater. The study producing 12 data points conducted tensile load testing to failure on 12 PMHS head and torso specimens where the skin and musculature were intact and T8-T11 were potted and secured to the base of the loading apparatus (Yliniemi et al., 2009). Eight males and four females were tested using aviation specific tensile loading rates ranging from 520 mm/s to 740 mm/s (Yliniemi et al., 2009). Mean subject anthropometry included the following: age (50.1 yrs), height (173.5 cm), and body mass (76.7 kg) (Yliniemi et al., 2009). Failure loads were recorded as well as detailed cervical spine injury from post-test radiographs using current AIS values. All subjects experienced AIS 3 or greater neck injury, and the mean tensile load at failure was 3100 N (3250 N for males, 2803 N for females). The study producing six data points was a whole PMHS frontal impact study where a combination of tension and flexion occurred in the subjects and injury was observed by post test autopsy (Cheng et al., 1982). Cervical spine tension of PMHS was calculated using observed accelerations and head/neck mass properties. Three of the data

points came from a study where pure axial tension was applied to the cervical spine of whole cadavers; loads were recorded and injury was specified (Yoganandan et al., 1996). The study that produced the remaining single data point was the only isolated spinal column tested and no injury was observed (Sances et al., 1981). **Table 21** provides a summary of the failure loads and anthropometry of all PMHS.

**Table 21. MANIC(Gz) Post Mortem Human Subject Peak Tensile Neck Load and Anthropometry**

| Sex | Type | Age (year) | Body Mass (kg) | Failure Load (N) | AIS 3+ Injury | Source |
|---|---|---|---|---|---|---|
| **M** | Whole | 66 | 72.5 | 3490 | Yes | Cheng et al., 1982 |
| **F** | Whole | 54 | 50 | 7200 | Yes | " |
| **M** | Whole | 56 | 96 | 2420 | Yes | " |
| **M** | Whole | 63 | 72.5 | 850 | No | " |
| **M** | Whole | 68 | 93 | 6520 | Yes | " |
| **M** | Whole | 67 | 60 | 3210 | No | " |
| **N/A** | Whole | 66±11 | N/A | 2400 | Yes | Yoganandan et al., 1996 |
| **N/A** | Whole | 66±11 | N/A | 3900 | Yes | " |
| **N/A** | Whole | 67 | 67 | 3800 | Yes | " |
| **N/A** | Isolated | 61 | 70 | 2688 | No | Sances et al., 1981 |
| **F** | Torso | 48 | 55 | 3560 | Yes | Yliniemi et al., 2009 |
| **F** | Torso | 45 | 59 | 2250 | Yes | " |
| **F** | Torso | 56 | 68 | 1910 | Yes | " |
| **F** | Torso | 43 | 74 | 3490 | Yes | " |
| **M** | Torso | 35 | 59 | 4060 | Yes | " |
| **M** | Torso | 48 | 73 | 3860 | Yes | " |
| **M** | Torso | 50 | 68 | 2810 | Yes | " |
| **M** | Torso | 60 | 77 | 3150 | Yes | " |
| **M** | Torso | 59 | 82 | 3230 | Yes | " |
| **M** | Torso | 37 | 77 | 3220 | Yes | " |
| **M** | Torso | 59 | 114 | 2440 | Yes | " |
| **M** | Torso | 61 | 114 | 3230 | Yes | " |

Risk functions were constructed using survival analysis for AIS 2+ and 3+ injury levels after methods used in research by Bass et al. and Parr et al. (Bass et al., 2006; Parr et al., 2013).

## Results

### MANIC(-Gx) Results

The individual AIS 2+ and 3+ risk functions and 95% confidence intervals generated for the –Gx axis of acceleration are provided in **Figure 29** and **Figure 30**. **Figure 31** shows both AIS 2+ and 3+ MANIC(Gx) risk functions on the same plot. The MANIC(Gx) risk functions were constructed by combining data from 67 human subjects from a single frontal impact experimental setup with data from a study with six PMHS. The equations of the AIS 2+ and 3+ risk functions are provided in Equation 15 and Equation 16. As stated previously the difference observed in the AIS 2+ and 3+ risk curves is produced by a single injury data point in the source data, indicating the sensitivity of the injury criteria when the sample size for the PMHS is small, as in this data set. These curves behave as would be expected. At the higher injury level, a greater value for MANIC(Gx) is allowed at a specific risk level. For example, at 5% risk of injury, the AIS 2+ risk curve allows for an MANIC(Gx)=0.56 and the AIS 3+ risk curve allows for an MANIC(Gx)=0.72. Table 22 summarizes the MANIC(Gx) predicted values at 5%, 10%, and 20% injury risk prediction levels, which are common thresholds in aviation and automotive safety applications: the associated 95% confidence intervals are also provided.

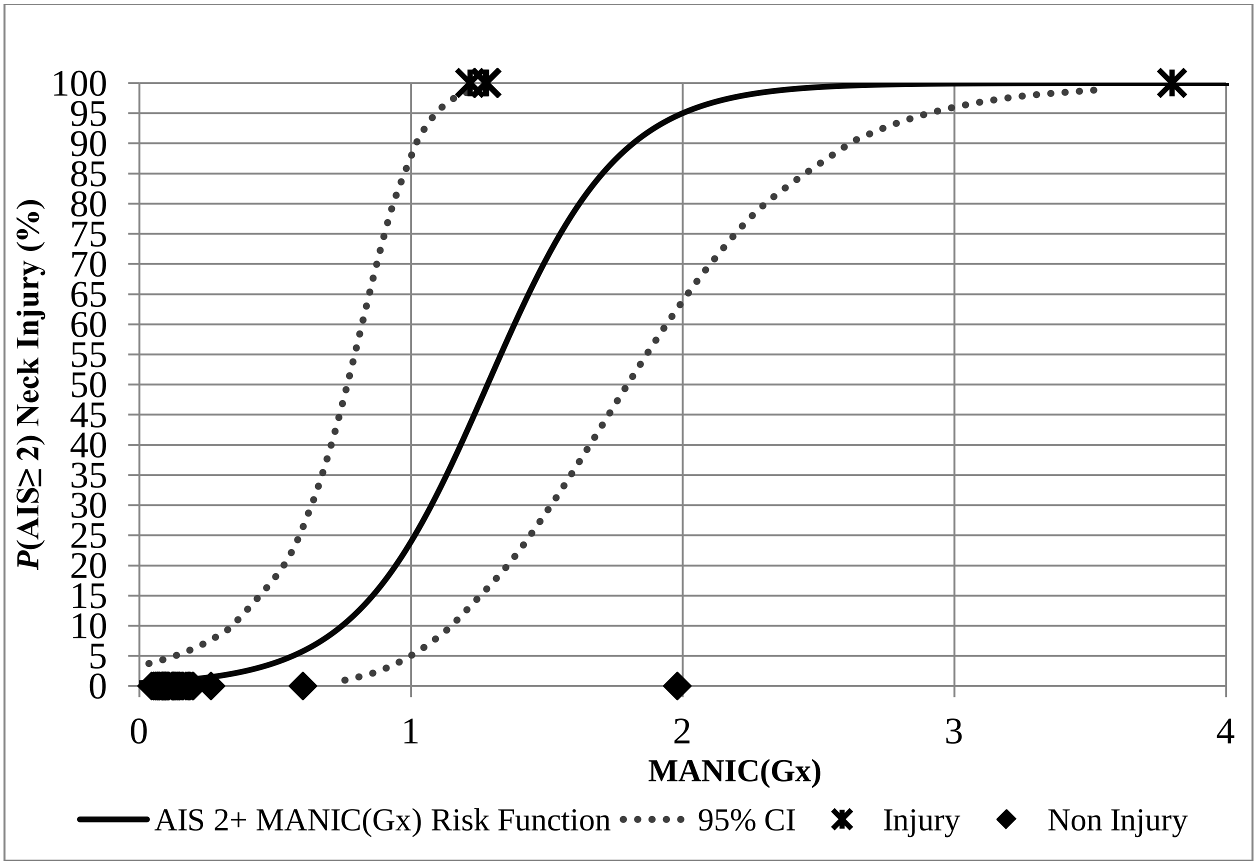


**Figure 29. MANIC(Gx) AIS 2+ Risk Function**

$$P(\text{AIS} \geq 2) = \frac{1}{1 + e^{5.2545 - 4.1 * \text{MANIC(Gx)}}} \tag{15}$$

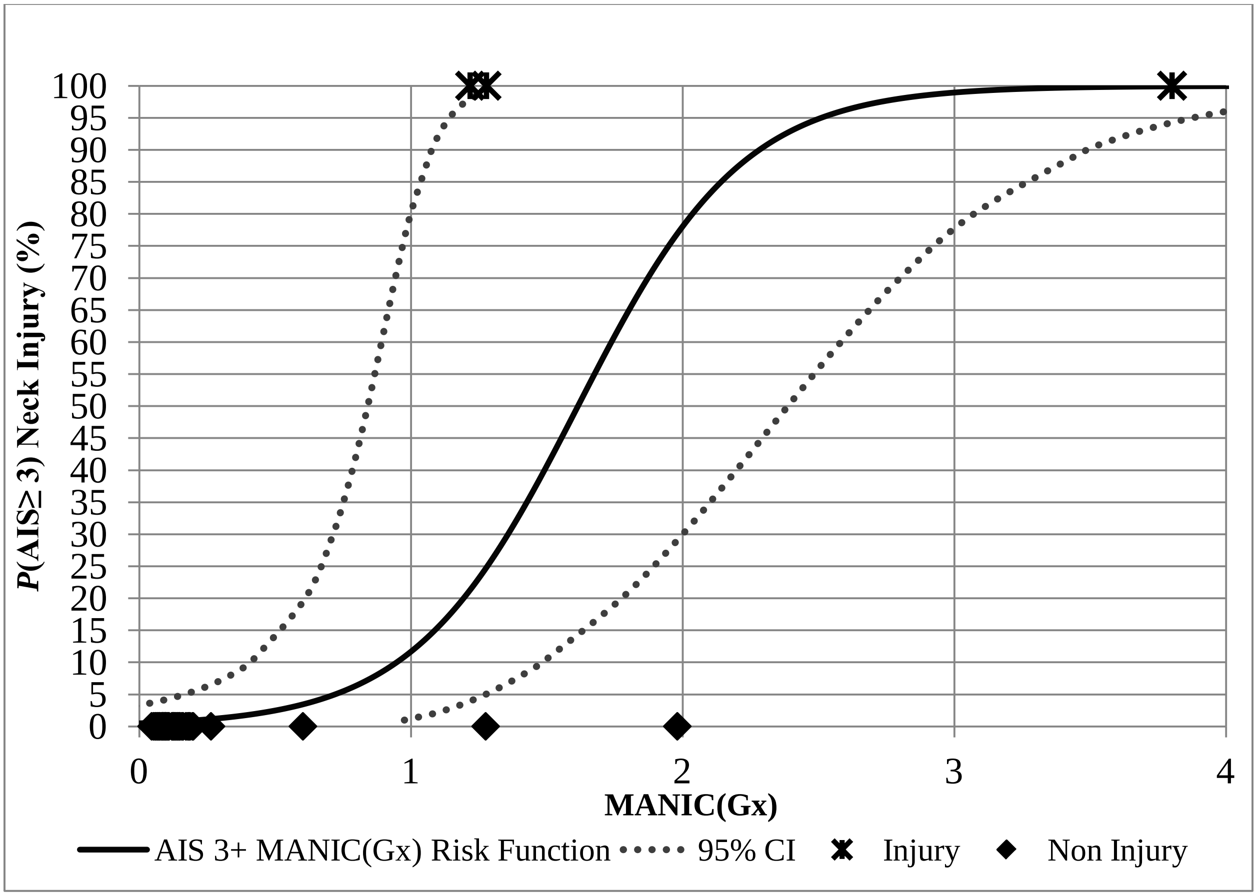


**Figure 30. MANIC(Gx) AIS 3+ Risk Function**

$$P(AIS \geq 3) = \frac{1}{1 + e^{5.31423 - 3.3922 * MANIC(Gx)}} \qquad (16)$$

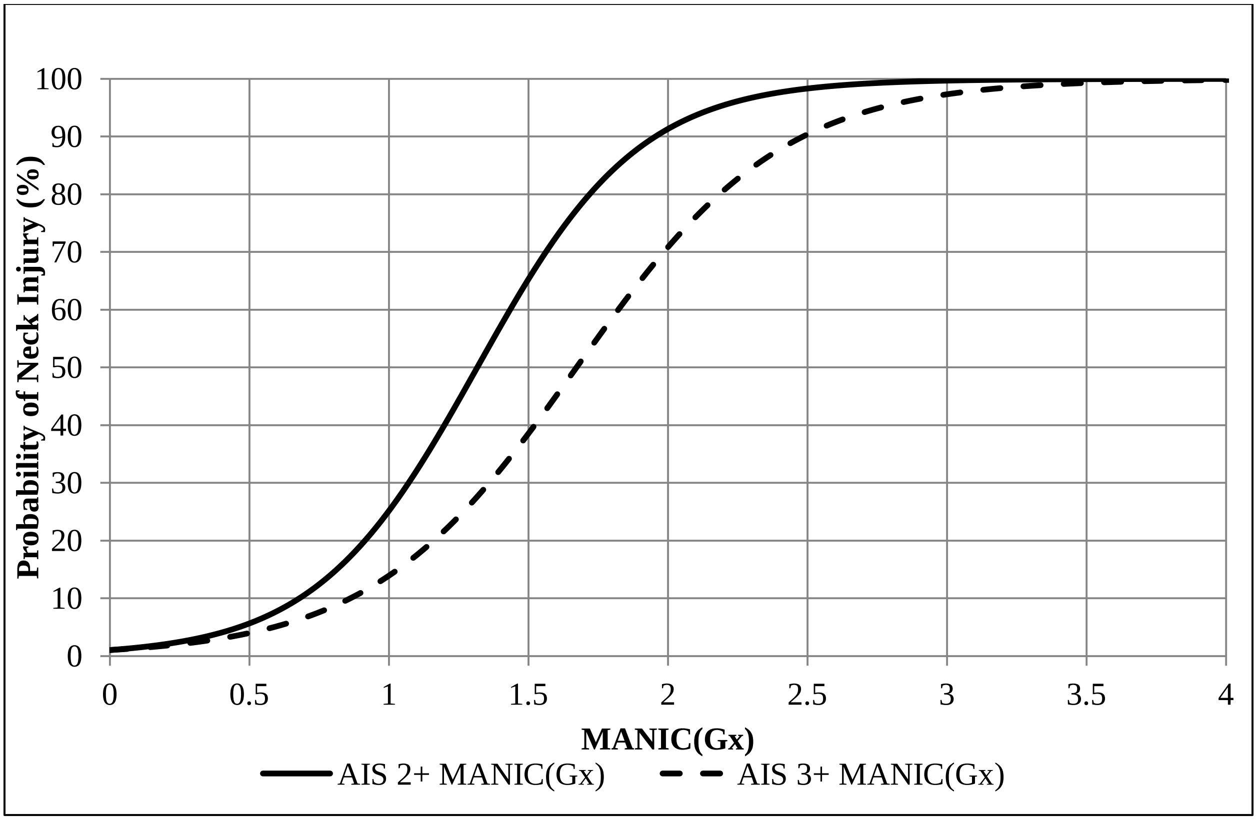


**Figure 31. Probability of AIS 2+ and AIS 3+ MANIC(Gx) Risk Functions**

**Table 22. MANIC(Gx) Summary of Predicted Values (95% Confidence Intervals)**

| | **AIS 2+ MANIC(Gx)** | **AIS 3+ MANIC(Gx)** |
|---|---|---|
| **5%** | 0.56 (0.13, 1.00) | 0.72 (0.17, 1.27) |
| **10%** | 0.75 (0.34. 1.15) | 0.95 (0.41, 1.48) |
| **20%** | 0.94 (0.53, 1.36) | 1.19 (0.61, 1.78) |

### MANIC(Gy) Results

MANIC(Gy) risk functions were constructed with SA using a combination of human subject and PMHS data to predict risk of AIS 2 or greater and AIS 3 or greater injury at a given MANIC(Gy) neck load. The AIS 2 + risk function is shown in **Figure 32** and the AIS 3+ risk

function is shown in **Figure 33**. For the AIS 2+ risk function data set, the MANIC(Gy) mean and standard deviation of the injurious data points was 1.07 and 0.71 respectively. The MANIC(Gy) mean and standard deviation of the AIS 2+ risk function non-injurious data points was 0.37 and 0.14 respectively. For the AIS 3+ risk function data set, the MANIC(Gy) mean and standard deviation of the injurious data points was 1.12 and 0.80 respectively. The MANIC(Gy) mean and standard deviation of the AIS 3+ risk function non-injurious data points was 0.38 and 0.15 respectively. The non-injury and injury data points are plotted at the location of their MANIC(Gy) values (x-axis) and at y-values of 0 or 100% respectively. Five data points were classified injurious at a level of AIS 2 or greater and 60 data points were non-injurious. A comparison of the AIS 2+ risk curve and the AIS 3+ risk curve is provided in **Figure 34**. The AIS 3+ risk function differs from the AIS 2+ risk function by a single PMHS data point which had an observed AIS 2 neck injury which was considered an injury data point for the data set used to produce the AIS 2+ risk function but was classified in the non-injurious category for the data set used to produce the AIS 3+ risk function. Thus, for the AIS 3+ risk function, four data points were injurious at a level of AIS 3+ and 61 data points were non-injurious. The difference between the AIS 2+ and AIS 3+ risk curves is produced by a single injury data point, indicating the sensitivity of the injury criteria when the PMHS injury data sample size is small, as it is in the current data set. The AIS 2+ risk function is provided below in Equation 17 and the AIS3+ risk function is provided in Equation 18.

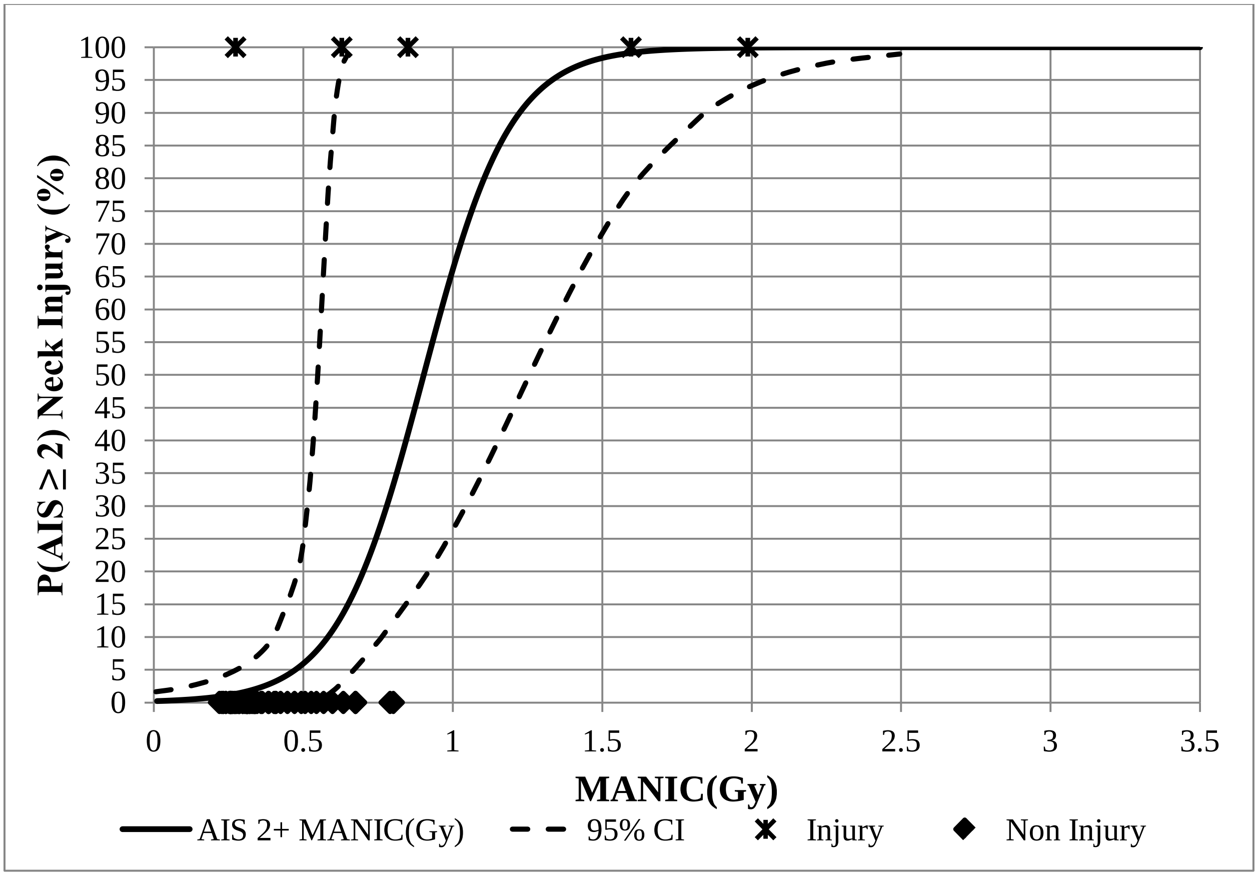


**Figure 32: Probability of AIS 2+ MANIC(Gy) Risk Function with 95% CI**

$$P(\mathrm{AIS} \geq 2) = \frac{1}{1 + e^{6.185 - 6.85 * \mathrm{MANIC(Gy)}}} \tag{17}$$

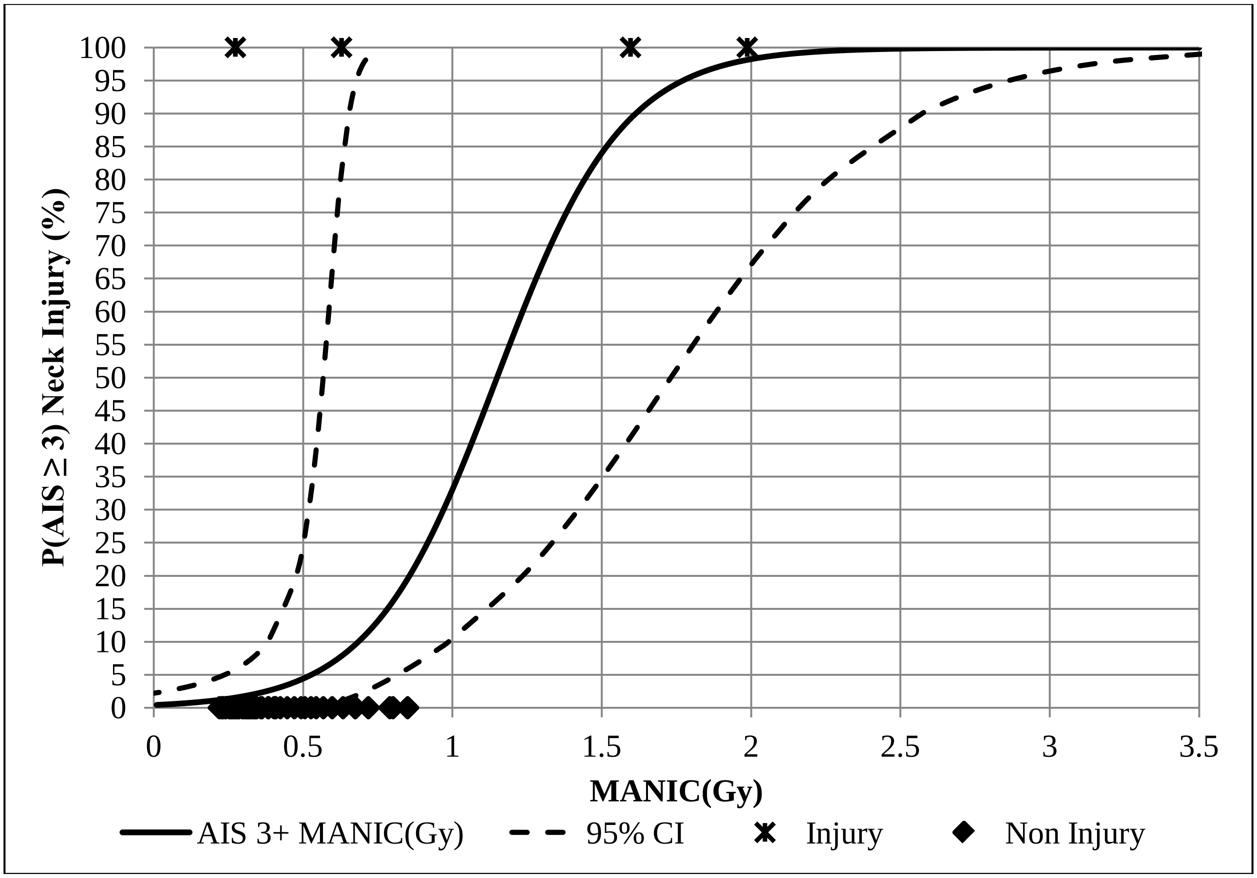


**Figure 33. Probability of AIS 3+ MANIC(Gy) Risk Function with 95% CI**

$$P(\text{AIS} \geq 3) = \frac{1}{1 + e^{5.44 - 4.73 * \text{MANIC(Gy)}}} \tag{18}$$

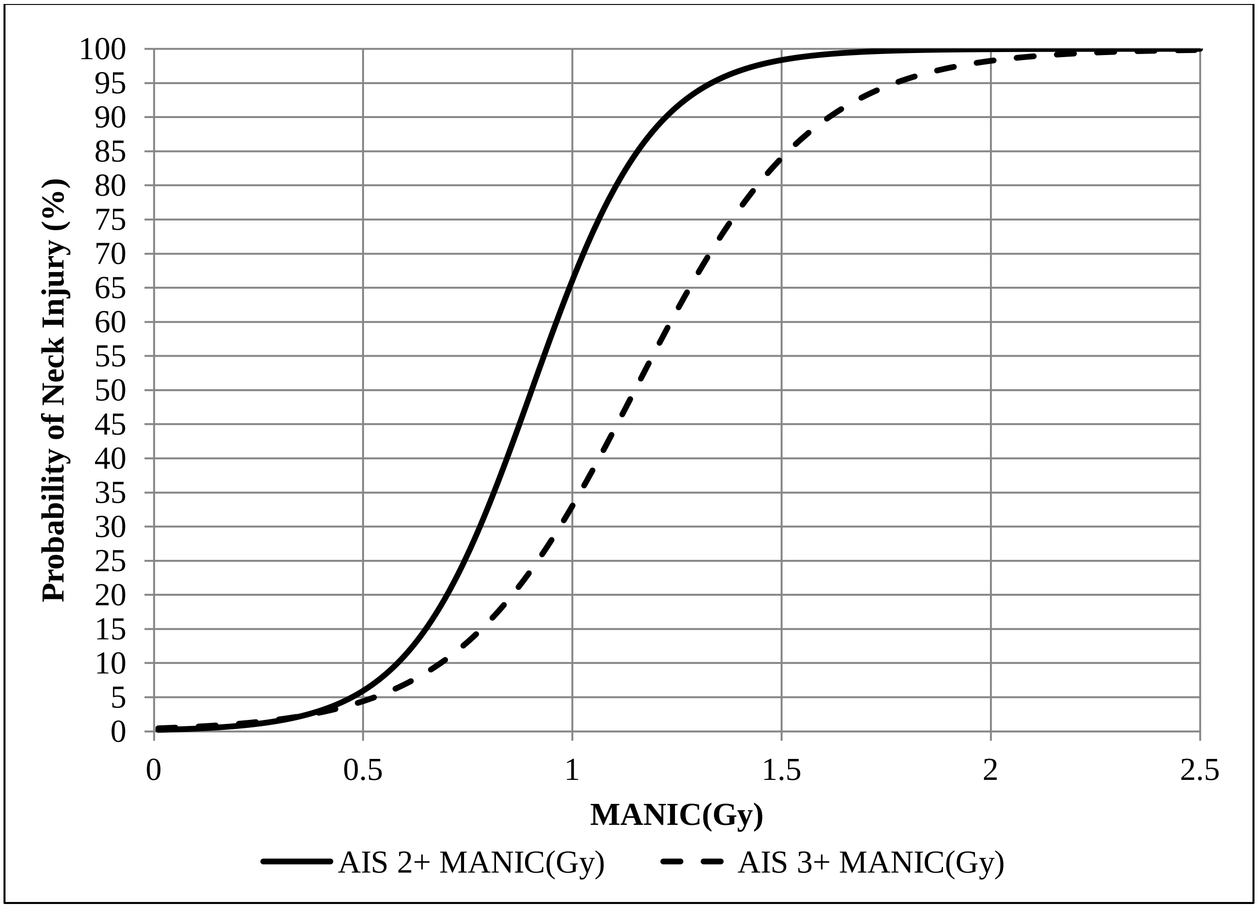


**Figure 34. MANIC(Gy) AIS 2+ and 3+ Risk Functions**

The AIS+ and AIS 3+ risk functions in **Figure 34** behave as expected. A greater value for MANIC(Gy) is allowed at a specific risk level at the higher injury level. For example, at 5% risk of injury, the AIS 2+ risk curve allows for a MANIC(Gy) = 0.473 and the AIS 3+ risk curve allows for a MANIC(Gy) = 0.527. Larger differences are observed at higher risk percentages as the two risk curves diverge between MANIC(Gy) values of 0.5 and 2.0. **Table 25** summarizes the MANIC(Gy) predicted values at commonly used injury risk percentages and the associated 95% confidence intervals.

Table 23. MANIC(Gy) Summary of Predicted Values (95% Confidence Intervals)

| | AIS 2+ MANIC(Gy) | AIS 3+ MANIC(Gy) |
|---|---|---|
| **5%** | 0.47 (0.27, 0.67) | 0.52 (0.24, 0.82) |
| **10%** | 0.58 (0.40. 0.76) | 0.68 (0.38, 0.99) |
| **20%** | 0.70 (0.48, 0.92) | 0.86 (0.48, 1.24) |

### MANIC(-Gz) Results

The AIS 2 and 3 or greater risk functions generated with the combined human subject and PMHS data are shown in **Figure 37, Figure 36,** and **Figure 37**. The 5% predicted tensile loads are 922 and 1136 N respectively. Equation 19 and Equation 20 provide the specific equations for the AIS 2+ and AIS 3+ risk functions respectively.

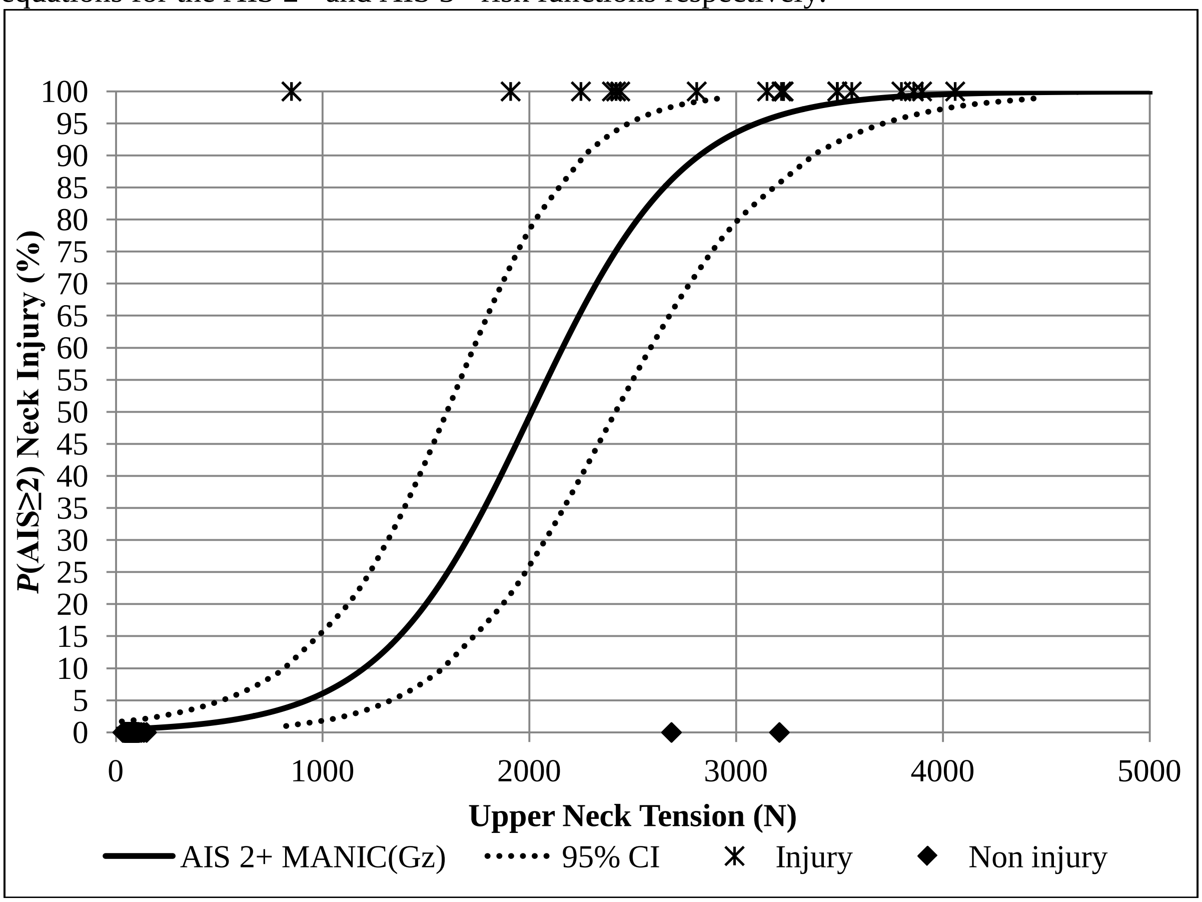


**Figure 35. MANIC(Gz) AIS 2+ Risk Function**

$$P(AIS \geq 2) = \frac{1}{1 + e^{5.44 - 0.00271 * MANIC(Gz)}} \tag{19}$$

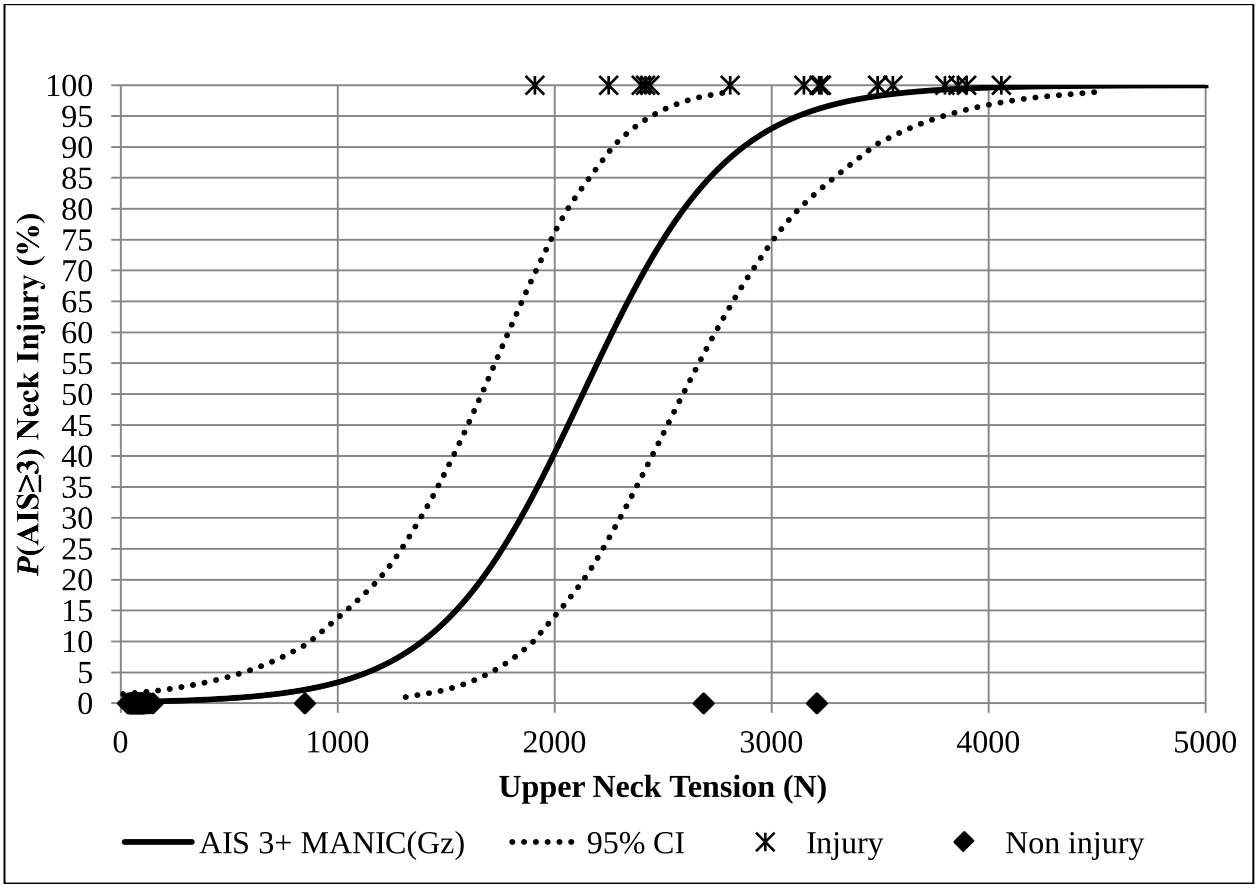


**Figure 36. MANIC(Gz) AIS 3+ Risk Function**

$$P(AIS \geq 3) = \frac{1}{1 + e^{6.318 - 0.00297 * MANIC(Gz)}} \tag{20}$$

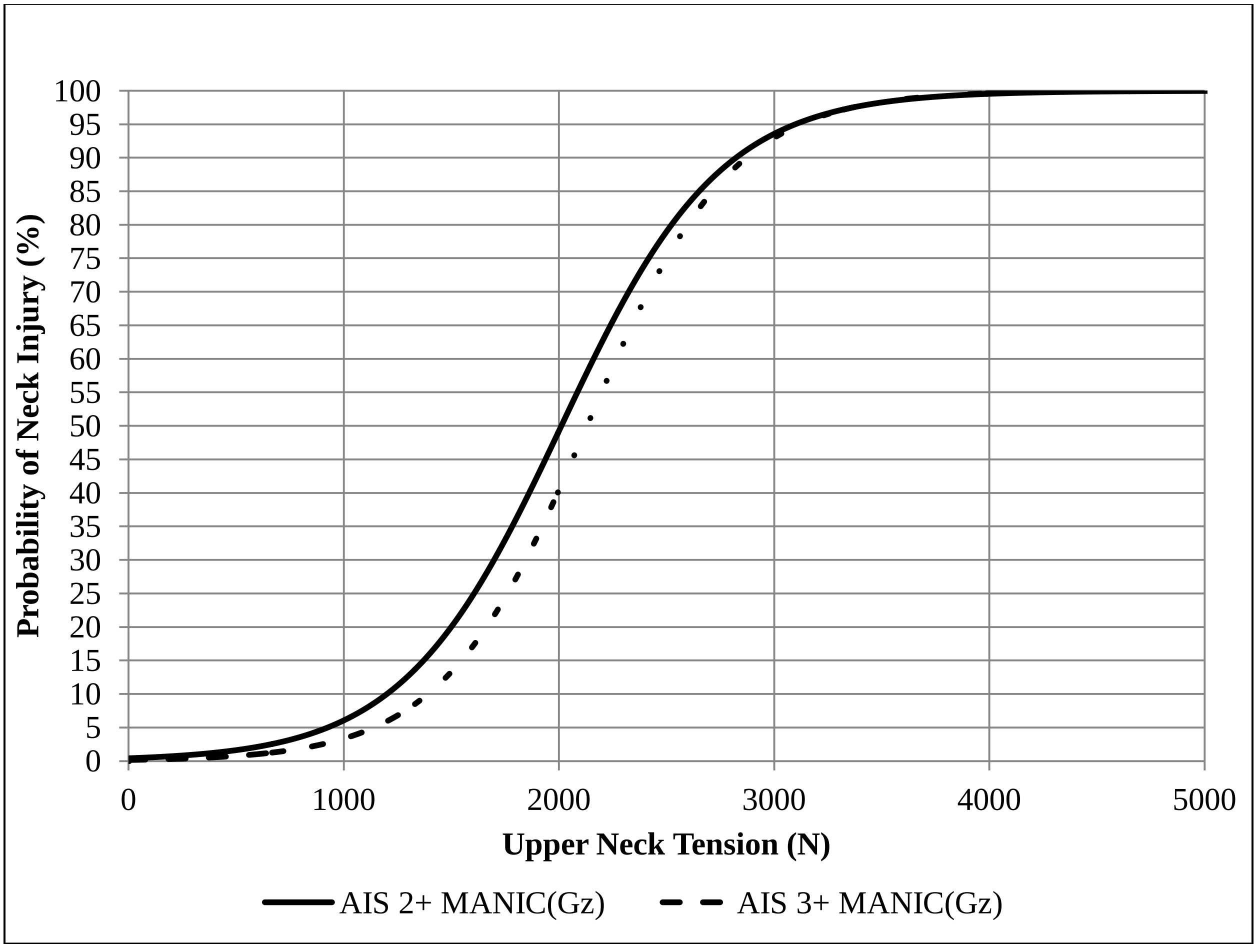


**Figure 37. MANIC(Gz) AIS 2+ and 3+ Risk Functions**

At the outset of the present study it was hoped that the data would allow for two separate MANIC(Gz) curves to be generated based upon actual mass or gender data of the combined 22 data points. To analyze the effect of body mass, peak tension of subjects with mass greater than the sample mean of 76.7 kg ($N$ = 13) was compared with peak tension of subjects with mass less than 76.7 kg ($N$ = 5). Body mass was not reported for two subjects. To analyze the effect of gender, peak tension of female subjects ($N$ = 5) was compared with peak tension of male subjects ($N$ = 13). Gender was not reported for four subjects. Based on this data set, neither body mass nor gender was a significant predictor of tensile neck loading using the Mann-Whitney U non-parametric test for body mass ($\alpha$= 0.05, $p$ = 0.72) and gender ($\alpha$ =0.05, $p$ = 0.84). Therefore, the

data was pooled and a single risk function for all body masses was created. **Table 24** provides a summary of the MANIC(Gz) predicted values at commonly used injury levels and the associated 95% confidence intervals.

**Table 24. MANIC(Gz) Summary of Predicted Values (95% Confidence Intervals)**

| | AIS 2+ MANIC(Gz) (N) | AIS 3+ MANIC(Gz) (N) |
|---|---|---|
| **5%** | 922 (513, 1330) | 1136 (566, 1707) |
| **10%** | 1198 (815. 1580) | 1388 (874, 1902) |
| **20%** | 1497 (1123, 1871) | 1661 (1188, 2135) |

### Summary of Results

**Table 25** provides a summary of the three axis-specific sub-elements that comprise the complete MANIC. Taken as a whole, the three subcomponents of the MANIC provide a family of preliminary, pilot scale, human based risk functions for injury risk protection at the 5% risk of AIS2+ and 3+ injury. The MANIC can be applied at various injury levels and injury risk percentages as desired by the practitioner. The next section provides an overview of how these criteria might be applied to sample escape system testing data sets.

**Table 25. MANIC Summary**

| Criteria Element | Limit |
|---|---|
| $MANIC(-Gx) = \left\lvert \frac{F_Z}{F_{Zcrit}} \right\rvert + \left\lvert \frac{M_Y}{M_{Ycrit}} \right\rvert$ | **Peak MANIC(-Gx) < 0.56**<br>Less than 5% Risk of AIS 2+ Injury<br>**(<0.72 for AIS 3+)** |
| $MANIC(Gy) = \sqrt{\left(\frac{F_X}{F_{Xcrit}}\right)^2 + \left(\frac{F_Y}{F_{Ycrit}}\right)^2 + \left(\frac{F_Z}{F_{Zcrit}}\right)^2 + \left(\frac{M_Y}{M_{Ycrit}}\right)^2 + \left(\frac{M_Z}{M_{Zcrit}}\right)^2}$ | **Peak MANIC(Gy) < 0.48**<br>Less than 5% Risk of AIS 2+ Injury<br>**(<0.53 for AIS 3+)** |
| $MANIC(-Gz) = +Fz$ | **Peak MANIC(-Gz) < 922 N/207 lb**<br>Less than 5% Risk of AIS 2+ Injury<br>**(<1136 N/256 lb for AIS 3+)** |

**Figure 38** illustrates the MANIC application process. In escape system qualification testing the ATD is subjected to the dynamic accelerative loading experienced during the full sequence of ejection. The ATD's primary neck loads are observed throughout the duration of the ejection. This data is used to compute a time history for each of the three MANIC subcomponents and the peak MANIC(Gx), MANIC(Gy), and MANIC(Gz) are determined. These peaks are compared to the maximum allowable MANIC(Gx), MANIC(Gy), and MANIC(Gz) values determined by the user. In the aviation specific MANIC application process depicted by the bottom section of **Figure 38** the user defined limits are a 5% risk of AIS 2+ neck injury. The specific load limit for each axis specific subcomponent is shown.

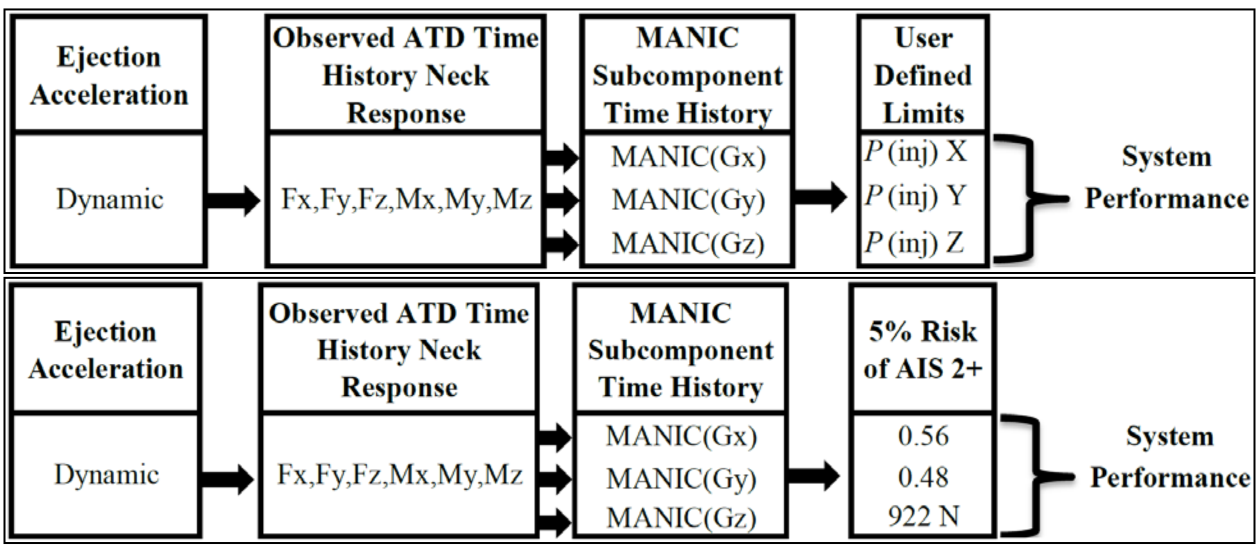


**Figure 38. Generic MANIC Application Process Compared (Top) to Sample Aviation Specific MANIC Application Process (Bottom)**

### MANIC Application Feasibility Analysis

To compare the performance of the complete pilot scale MANIC to a legacy neck injury criteria, the MANIC and the NIC were applied to two different ejection data sets. These data

sets were full sequence escape system tests with aerospace ATDs where the six primary upper neck loads were observed for the entire time history of the ejection. The Knots Equivalent Air Speed (KEAS), ATD nude mass, MANIC sub-criteria information, and NIC information are presented in the following tables. For ATD nude mass of 96 to 135 lbs the Small Female Hybrid III neck was used, for ATD nude mass of 136 to 199 lbs the Mid Male Hybrid III neck was used, and for ATD nude mass of 200 to 245 lbs the Large Male Hybrid III neck was used (Nichols, 2006). To calculate the NIC, the 103 lb ATD critical values from **Table 20** were used for ATD mass of 103 to 144 lbs, the 150 lb ATD critical values from **Table 20** were used for ATD mass of 145 lbs, and the 220 lb ATD critical values from **Table 20** were used for ATD mass of 245 lbs. Upon test initiation, the rocket sled brings the ejection system to the desired speed on the test track. Then the seat fires and the ejection system operates to include catapult, drogue parachute deployment, main parachute deployment, man/seat separation, and concluding with the parachute phase and touchdown. This entire sequence ranges from approximately four to 15 seconds from start to finish depending on ATD mass, sled speed, and wind conditions.

The application of the laboratory-produced, human-based MANIC to the full sequence, ATD, ejection data requires that some assumptions be made. One assumption is that the ATD neck loads are representative of a human. Some studies have shown this to be problematic in specific modes of loading. For example, Buhrman and Perry found that the Hybrid III neck in the Advanced Dynamic Anthropomorphic Manikin (ADAM) which they used in vertical accelerative laboratory ejection testing was not precisely representative of human response in both neck shear force and bending moment (Buhrman and Perry, 1994). Bass et al. detailed the kinematic differences in matched pair experiments with PMHS and ATDs using the Hybrid III neck using high speed camera images (Bass et al., 2006). These types of studies raise questions about the applicability and sensitivity of the Hybrid III neck to be representative of human neck

loads. At the same time, however, human test subjects cannot be exposed to forces which have the potential to cause irreparable harm, creating a paradox for the application of robust human-based criteria from which to adequately evaluate the safety of a system. Since investigating the need for a human to ATD transfer function and developing the transfer function if warranted is outside the scope of the present research, the MANIC is directly applied to Hybrid III ATD neck load data as-is as a preliminary step in the evaluation of the MANIC's performance with real world data. If a human to ATD transfer function or a biofidelic ATD becomes available, the MANIC the application of the MANIC can be applied with improved confidence. Another assumption is that the injury risk functions constructed with data from single axis, laboratory accelerative experiments capture the neck loading injury mechanisms and thus are appropriate for application to the dynamic escape environment.

Part of applying the NIC is the involvement of a panel of subject matter experts (SMEs) that review the data if any of the 12 criteria are exceeded (Nichols, 2006). According to Nichols, the event(s) causing the limit exceedences are to be investigated, to include factors such as "body position; off axis neck loading; seat, chest, and head linear and angular acceleration; the portion of the limit curve that was exceeded; and the magnitude of the exceedence (Nichols, 2006)." He points out further that some exceedences may be perfectly fine, while others may not pose risk depending on the factors observed in the ATD at the time of the test and the other loads occurring. This is problematic, especially when administering criteria during developmental testing which is a critical phase in the acquisition lifecycle. The program and the contractor require timely feedback about success or failure of the test. In escape system testing, evaluating reliability is a primary component of the testing as is system safety. A set of neck injury criteria is of questionable benefit if the criteria cannot accurately provide feedback on system safety in the form of predicted risk of injury resulting from the test. Allowing some subjective evaluation

of the data and review of the criteria results may indeed be warranted, but with no validated risk functions for any of the 12 sub-criteria, the SMEs analyzing NIC data do not have adequate risk prediction tools from which to make their recommendations.

Since nine of the 12 sub-elements of the NIC do not have specified injury levels associated with the limits, performance of the NIC is compared to performance of both AIS 2+ and AIS 3+ MANIC limits (NIC upper/lower neck tension duration and upper neck Nij limits are loosely tied to 10% risk of AIS 3+ neck injury). **Table 26** shows the results of the MANIC evaluated at 5% risk of AIS 2+ injury with details provided for each of the three subcomponent criteria compared to the results of the NIC. It should be noted that this preliminary assessment of the MANIC performance using real-world data is designed to show how a fully developed MANIC might be implemented if accepted as the criteria for future developmental testing evaluations. The application of this initial pilot scale MANIC is provided to demonstrate the process and to compare the results for this metric to the NIC. Additional human subject and PMHS experimental data would be required to further develop robust MANIC, risk functions.

**Table 26. Data Set 1 – AIS 2+ MANIC 5% Results Compared to NIC Results**

| | | | MANIC Results | | | NIC Results |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.56= 5% AIS2+)** | **Gy (<0.48= 5% AIS2+)** | **Gz (<922N= 5% AIS3+)** | |
| **1** | 442 | 145 | Pass (0.47, 3.5%) | Pass (0.43, 3.6%) | Fail (2134.5N, 58.4%) | Pass |
| **2** | 227 | 145 | Fail (0.76, 10.5%) | Fail (0.68, 18.26%) | Fail (3122.9N, 95.3%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **3** | 237 | 145 | Pass (0.32, 1.9%) | Pass (0.478, 4.98%) | Fail (1464.8N, 18.6%) | Pass |
| **4** | 247 | 145 | Pass 0.48, | Fail (0.71, | Fail (2112.4N, | Pass |

| | | | | | | |
|---|---|---|---|---|---|---|
| | | | 3.6%) | 21.5%) | 57.0%) | |
| **5** | 0 | 245 | Pass (0.19, 1.1%) | Pass (0.39, 2.8%) | Pass (913.1N, 4.9%) | Pass |
| **6** | 444 | 245 | Pass (0.38, 2.4%) | Pass (0.42, 3.32%) | Fail (2540.2N, 80.8%) | Pass |
| **7** | 0 | 138 | Pass (0.39, 2.5 %) | Pass (0.41, 3.1%) | Fail (1657.7N, 27.9%) | Pass |
| **8** | 0 | 140 | Pass (0.28, 1.6%) | Pass (0.43, 3.5%) | Pass (868.6N, 4.4%) | Pass |
| **9** | 143 | 245 | Pass (0.29, 1.7%) | Pass (0.29, 1.4%) | Fail (1415.6N, 16.7% | Pass |
| **10** | 434 | 245 | Pass (0.52, 4.2%) | Fail (0.52, 6.36%) | Fail (3489.2N, 98.3%) | Fail 3/12. **Passed after SME review.** |
| **11** | 429 | 137 | Fail (0.71, 8.8%) | Fail (0.69, 18.7%) | Fail (3363.2N, 97.5%) | Fail 7/12 (UN & LN tension & shear duration, UN & LN Nij, UNMIx). **Failed after SME review.** |
| **12** | 439 | 140 | Fail (0.70, 8.4%) | Fail (0.59, 10.33%) | Fail (2687.9N, 86.3%) | Fail 4/12 (tension & shear duration, Nij, UNMIx). **Passed after SME review.** |
| **13** | 235 | 127 | Fail (0.70, 8.4%) | Fail (0.82, 43.5%) | Fail (2821.6N, 90.3%) | Fail 2/12 (tension duration & Nij). **Failed after SME review.** |
| **14** | 235 | 127 | Pass (0.42, 2.8%) | Fail (0.53, 7.02%) | Fail (1594.6N, 24.6%) | Pass |
| **15** | 0 | 245 | Fail (0.86, 15.1%) | Fail (0.66, 15.9%) | Pass (276.9N, 0.91%) | Failure (Not reported due to seat malfunction) |
| **16** | 0 | 245 | Pass (0.36, 2.2%) | Pass (0.43, 3.6%) | Pass (253.4N, 0.85%) | Pass |
| **17** | 222 | 103 | Pass (0.36, 2.2%) | Pass (0.476, 4.9%) | Fail (1305.6N, 12.9%) | Pass |
| **18** | 452 | 103 | Fail (0.64, 6.6%) | Fail (0.58, 9.7%) | Fail (2000.3N, 49.4%) | Fail 2/12 (UN & LN tension duration). **Passed after SME** |

| | | | | | | |
|---|---|---|---|---|---|---|
| | | | | | | **review.** |
| **19** | 452 | 103 | Pass (0.49, 3.8%) | Fail (0.56, 8.5%) | Fail (1687.3N, 29.5%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **20** | 530 | 245 | Pass (0.40, 2.6%) | Pass (0.47, 4.7%) | Fail (1384.6N, 15.6%) | Fail 1/12 (LN shear duration). **Passed after SME review.** |
| **21** | 575 | 245 | Pass (0.27, 1.5%) | Pass (0.42, 3.4%) | Fail (1062.0N, 7.1%) | Pass |
| **22** | 338 | 245 | Pass (0.39, 2.5%) | Pass (0.47, 4.7%) | Fail (1249.4N, 11.3%) | Pass |
| **23** | 0 | 103 | Pass (0.39, 2.5%) | Pass (0.44, 3.9%) | Pass (859.3N, 4.3%) | Pass |
| **24** | 155 | 245 | Pass (0.32, 1.9%) | Pass (0.35, 2.1%) | Pass (869.5N, 4.4%) | Pass |
| **25** | 162 | 245 | Pass (0.25, 1.4%) | Pass (0.4, 2.9%) | Fail (1156.05N, 9.0%) | Pass |
| **26** | 171 | 245 | Pass (0.34, 2.1%) | Pass (0.34, 1.9%) | Pass (770.6N, 3.4%) | Pass |
| **27** | 437 | 245 | Pass (0.29, 1.7%) | Pass (0.38, 2.6%) | Fail (2182N, 61.5%) | Pass |
| **28** | 542 | 103 | Fail (0.75, 10.2%) | Fail (0.75, 26.2%) | Fail (2388.8N, 73.7%) | Fail 3/12 (UN & LN tension duration, UN Nij) |
| **29** | 260 | 136 | Pass (0.28, 1.6%) | Pass (0.38, 2.6%) | Fail (1221.8N, 10.6%) | Pass |
| **30** | 161 | 103 | Pass (0.28, 1.6%) | Pass (0.43, 3.6%) | Pass (518.8N, 1.7%) | Pass |
| **31** | 152 | 138 | Pass (0.34, 2.1%) | Pass (0.39, 2.7%) | Fail (1080.8N, 7.5%) | Pass |
| **32** | 225 | 103 | Pass (0.44, 3.1%) | Pass (0.4, 2.9%) | Fail (1601.7N, 25.0%) | Pass |
| **33** | 0 | 103 | Pass (0.54, 4.6%) | Fail (0.7, 20.0%) | Fail (1251.8N, 11.4%) | Fail 1/12 (LN compression duration). **Passed after SME review.** |

Comparing the results of the MANIC with the NIC from the AIS 2+ Data Set 1 in **Table 26**, 26 of the 33 tests failed one or more elements of the 5% AIS 2+ MANIC, while pre-SME review 11 of the 33 failed one or more elements of the NIC (the SMEs reversed seven tests from fail to pass for a total of 29 passes). The 5% AIS 2+ MANIC agreed with the NIC on seven of the 22 pre-SME passes and agreed with the NIC on 11 of 11 pre-SME failures. Overall, 22 of the 33 tests passed the pre-SME NIC while seven of the 33 tests passed the 5% AIS 2+ MANIC. There were 15 tests which failed one or more element of the 5% AIS 2+ MANIC which passed the pre-SME NIC. All 15 of these 5% AIS 2+ MANIC failures included MANIC(Gz) and two of 15 included failure of MANIC(Gy). The MANIC evaluated at the 5% risk AIS 2+ injury is more conservative than the NIC (which allows for AIS3+ injuries in most elements), but it does provide insight into the specific risk posed by the observed neck loads

**Table 27** compares the MANIC evaluated at 5% risk of AIS 3+ injury to the NIC. It was expected that the difference between the 5% AIS 2+ MANIC and the 5% AIS 3+ MANIC results would have been observable in the real world data and this indeed was the case. Overall, the 5% AIS 2+ MANIC was more conservative (allowed for less load at the same percent risk of injury) than the 5% AIS 3+ MANIC. Four tests failed the 5% AIS 2+ MANIC(Gx) limit but then passed the 5% AIS 3+ MANIC(Gx) limit. One test failed the 5% AIS 2+ MANIC(Gy) limit but passed the 5% AIS 3+ MANIC(Gy) limit. Two tests failed the 5% AIS 2+ MANIC(Gz) limit but passed the 5% AIS 3+ MANIC(Gz) limit. The two instances where the MANIC(Gz) reversed from fail to pass were cases where the MANIC(Gz) was the only MANIC sub element that failed, which resulted in the entire MANIC moving from fail to pass. Therefore the number of agreements and disagreements between the two criteria changed to MANIC / NIC agreement on 9 of 22 passes and 11 of 11 failures. These results demonstrate the sensitivity of the complete MANIC to specific AIS injury levels, a capability not available in the NIC.

Table 27. Data Set 1 – AIS 3+ MANIC 5% Results Compared to NIC Results

| | | | MANIC Results | | | NIC Results |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.72= 5% AIS3+)** | **Gy (<0.53= 5% AIS3+)** | **Gz (<1136N= 5% AIS3+)** | |
| **1** | 442 | 145 | Pass (0.47, 2.3%) | Pass (0.43, 3.2%) | Fail (2134.5N, 50.5%) | Pass |
| **2** | 227 | 145 | Fail (0.76, 5.7%) | Fail (0.68, 9.8%) | Fail (3122.9N, 95%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **3** | 237 | 145 | Pass (0.32, 1.4%) | Pass (0.478, 3.9%) | Fail (1464.8N, 12.3%) | Pass |
| **4** | 247 | 145 | Pass 0.48, 2.3%) | Fail (0.71, 11.1%) | Fail (2112.4N, 48.9%) | Pass |
| **5** | 0 | 245 | Pass (0.19, 0.9%) | Pass (0.39, 2.7%) | Pass (913.1N, 2.64%) | Pass |
| **6** | 444 | 245 | Pass (0.38, 1.7%) | Pass (0.42, 3.1%) | Fail (2540.2N, 77.3%) | Pass |
| **7** | 0 | 138 | Pass (0.39, 1.7%) | Pass (0.41, 2.9%) | Fail (1657.7N, 19.8%) | Pass |
| **8** | 0 | 140 | Pass (0.28, 1.2%) | Pass (0.43, 3.2%) | Pass (868.6N, 2.33%) | Pass |
| **9** | 143 | 245 | Pass (0.29, 1.3%) | Pass (0.29, 1.7%) | Fail (1415.6N, 10.8% | Pass |
| **10** | 434 | 245 | Pass (0.52, 2.7%) | Pass (0.52, 4.8%) | Fail (3489.2N, 98.3%) | Fail 3/12. **Passed after SME review.** |
| **11** | 429 | 137 | Pass (0.71, 4.8%) | Fail (0.69, 10.2%) | Fail (3363.2N, 97.5%) | Fail 7/12 (UN & LN tension & shear duration, UN & LN Nij, UNMIx). **Failed after SME review.** |
| **12** | 439 | 140 | Pass (0.70, 4.7%) | Fail (0.59, 6.6%) | Fail (2687.9N, 84.1%) | Fail 4/12 (tension & shear duration, Nij, UNMIx). **Passed after SME review.** |
| **13** | 235 | 127 | Pass (0.70, | Fail (0.82, | Fail (2821.6N, | Fail 2/12 (tension duration & Nij). |

| | | | | | | |
|---|---|---|---|---|---|---|
| | | | 4.7%) | 17.4%) | 88.7%) | **Failed after SME review.** |
| **14** | 235 | 127 | Pass (0.42, 1.9%) | Fail (0.53, 5.07%) | Fail (1594.6N, 17.04%) | Pass |
| **15** | 0 | 245 | Fail (0.86, 7.7%) | Fail (0.66, 9.0%) | Pass (276.9N, 0.4%) | Fail (Not reported due to sheared O-ring causing failure) |
| **16** | 0 | 245 | Pass (0.36, 1.6%) | Pass (0.43, 3.2%) | Pass (253.4N, 0.38%) | Pass |
| **17** | 222 | 103 | Pass (0.36, 1.6%) | Pass (0.476, 4.0%) | Fail (1305.6N, 8.01%) | Pass |
| **18** | 452 | 103 | Pass (0.64, 3.9%) | Fail (0.58, 6.3%) | Fail (2000.3N, 40.6%) | Fail 2/12 (UN & LN tension duration). **Passed after SME review.** |
| **19** | 452 | 103 | Pass (0.49, 2.4%) | Fail (0.56, 5.8%) | Fail (1687.3N, 21.3%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **20** | 530 | 245 | Pass (0.40, 1.8%) | Pass (0.47, 3.9%) | Fail (1384.6N, 9.9%) | Fail 1/12 (LN shear duration). **Passed after SME review.** |
| **21** | 575 | 245 | Pass (0.27, 1.2%) | Pass (0.42, 3.1%) | Pass (1062.0N, 4.1%) | Pass |
| **22** | 338 | 245 | Pass (0.39, 1.7%) | Pass (0.47, 3.8%) | Fail (1249.4N, 6.9%) | Pass |
| **23** | 0 | 103 | Pass (0.39, 1.7%) | Pass (0.44, 3.4%) | Pass (859.3N, 2.3%) | Pass |
| **24** | 155 | 245 | Pass (0.32, 1.4%) | Pass (0.35, 2.2%) | Pass (869.5N, 2.3%) | Pass |
| **25** | 162 | 245 | Pass (0.25, 1.1%) | Pass (0.4, 2.8%) | Fail (1156N, 5.3%) | Pass |
| **26** | 171 | 245 | Pass (0.34, 1.5%) | Pass (0.34, 2.1%) | Pass (770.6N, 1.7%) | Pass |
| **27** | 437 | 245 | Pass (0.29, 1.3%) | Pass (0.38, 2.6%) | Fail (2182N, 54.0%) | Pass |
| **28** | 542 | 103 | Fail (0.75, 5.5%) | Fail (0.75, 13.1%) | Fail (2388.8N, 68.5%) | Fail 3/12 (UN & LN tension duration, UN Nij) |

| | | | | | | |
|---|---|---|---|---|---|---|
| **29** | 260 | 136 | Pass (0.28, 1.2%) | Pass (0.38, 2.6%) | Fail (1221.8N, 6.3%) | Pass |
| **30** | 161 | 103 | Pass (0.28, 1.2%) | Pass (0.43, 3.2%) | Pass (518.8N, 0.83%) | Pass |
| **31** | 152 | 138 | Pass (0.34, 1.5%) | Pass (0.39, 2.7%) | Pass (1080.8N, 4.3%) | Pass |
| **32** | 225 | 103 | Pass (0.44, 2.1%) | Pass (0.4, 2.8%) | Fail (1601.7N, 17.3%) | Pass |
| **33** | 0 | 103 | Pass (0.54, 2.8%) | Fail (0.7, 10.7%) | Fail (1251.8N, 6.9%) | Fail 1/12 (LN compression duration). **Passed after SME review.** |

**Table 28** provides the most direct comparison of the MANIC and the NIC. It compares the MANIC evaluated at 10% risk of AIS 3+ injury to the NIC. In general the NIC is considered to be loosely tied to a 10% risk of AIS 3+ injury, though the accuracy of this criteria is not verifiable, thus varying levels of the MANIC have been compared to the NIC in this section. Two tests passed the MANIC(Gx) 10% AIS 3+ limit that failed the 5% AIS 3+ limit. Six tests passed the MANIC(Gy) 10% AIS 3+ limit that failed the 5% AIS 3+ limit. Six tests passed the MANIC(Gz) 10% AIS 3+ limit that failed the 5% AIS 3+ limit. Overall the MANIC sub element risk criteria behaved as expected as observed by application to real world data; the 5% AIS 3+ limits were more conservative than the 10% AIS 3+ limits. Four instances of reversal resulted in a reversal of the test conclusion from fail to pass, but one reversal also changed a previously agreed upon fail to pass the MANIC but remain failed in the NIC. The number of agreements and disagreements between the two criteria changed to MANIC / NIC agreement on 11 of 22 passes and 10 of 11 failures. These outcomes of comparing the most representative form of the MANIC to the NIC resulted in better agreement between the two criteria. However, there are still half of the tests that the MANIC failed due to the MANIC(Gz) limit which the NIC passed. The conservative MANIC(Gz) limit is based upon the most current tensile PMHS data

in the literature and as a single force criterion is not subject to the introduction of error by inaccurate critical values or structural issues from combined loading equations. Future work should evaluate if the MANIC(Gz) limits are overly conservative . If real world ejection injury results show that tensile injuries are not occurring as commonly as the MANIC(Gz) would predict, then modifications should be made to the MANIC(Gz). The reason for the numerous failures of the MANIC(Gz) tensile criteria may also be due to the direct application of the human MANIC to the ATD neck load data. The MANIC is being evaluated as-is in the present study, but a yet to be developed transfer function may adequately modify the MANIC(Gz) for more appropriate application to ATD neck loads.

**Table 28. Data Set 1 – AIS 3+ MANIC 10% Results Compared to NIC Results**

| | | | MANIC Results | | | NIC Results |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.95= 10% AIS3+)** | **Gy (<0.68= 10% AIS3+)** | **Gz (<1388N= 10% AIS3+)** | |
| **1** | 442 | 145 | Pass (0.47, 2.3%) | Pass (0.43, 3.2%) | Fail (2134.5N, 50.5%) | Pass |
| **2** | 227 | 145 | Pass (0.76, 5.7%) | Pass (0.68, 9.8%) | Fail (3122.9N, 95%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **3** | 237 | 145 | Pass (0.32, 1.4%) | Pass (0.478, 3.9%) | Fail (1464.8N, 12.3%) | Pass |
| **4** | 247 | 145 | Pass 0.48, 2.3%) | Fail (0.71, 11.1%) | Fail (2112.4N, 48.9%) | Pass |
| **5** | 0 | 245 | Pass (0.19, 0.9%) | Pass (0.39, 2.7%) | Pass (913.1N, 2.64%) | Pass |
| **6** | 444 | 245 | Pass (0.38, 1.7%) | Pass (0.42, 3.1%) | Fail (2540.2N, 77.3%) | Pass |
| **7** | 0 | 138 | Pass (0.39, 1.7%) | Pass (0.41, 2.9%) | Fail (1657.7N, 19.8%) | Pass |
| **8** | 0 | 140 | Pass (0.28, 1.2%) | Pass (0.43, 3.2%) | Pass (868.6N, 2.33%) | Pass |

| | | | | | | |
|---|---|---|---|---|---|---|
| **9** | 143 | 245 | Pass (0.29, 1.3%) | Pass (0.29, 1.7%) | Fail (1415.6N, 10.8% | Pass |
| **10** | 434 | 245 | Pass (0.52, 2.7%) | Pass (0.52, 4.8%) | Fail (3489.2N, 98.3%) | Fail 3/12. **Passed after SME review.** |
| **11** | 429 | 137 | Pass (0.71, 4.8%) | Fail (0.69, 10.2%) | Fail (3363.2N, 97.5%) | Fail 7/12 (UN & LN tension & shear duration, UN & LN Nij, UNMIx). **Failed after SME review.** |
| **12** | 439 | 140 | Pass (0.70, 4.7%) | Pass (0.59, 6.6%) | Fail (2687.9N, 84.1%) | Fail 4/12 (tension & shear duration, Nij, UNMIx). **Passed after SME review.** |
| **13** | 235 | 127 | Pass (0.70, 4.7%) | Fail (0.82, 17.4%) | Fail (2821.6N, 88.7%) | Fail 2/12 (tension duration & Nij). **Failed after SME review.** |
| **14** | 235 | 127 | Pass (0.42, 1.9%) | Pass (0.53, 5.07%) | Fail (1594.6N, 17.04%) | Pass |
| **15** | 0 | 245 | Pass (0.86, 7.7%) | Pass (0.66, 9.0%) | Pass (276.9N, 0.4%) | Fail (Not reported due to sheared O-ring causing failure) |
| **16** | 0 | 245 | Pass (0.36, 1.6%) | Pass (0.43, 3.2%) | Pass (253.4N, 0.38%) | Pass |
| **17** | 222 | 103 | Pass (0.36, 1.6%) | Pass (0.476, 4.0%) | Pass (1305.6N, 8.01%) | Pass |
| **18** | 452 | 103 | Pass (0.64, 3.9%) | Pass (0.58, 6.3%) | Fail (2000.3N, 40.6%) | Fail 2/12 (UN & LN tension duration). **Passed after SME review.** |
| **19** | 452 | 103 | Pass (0.49, 2.4%) | Pass (0.56, 5.8%) | Fail (1687.3N, 21.3%) | Fail 3/12 (UN & LN tension duration, Nij). **Passed after SME review.** |
| **20** | 530 | 245 | Pass (0.40, 1.8%) | Pass (0.47, 3.9%) | Pass (1384.6N, 9.9%) | Fail 1/12 (LN shear duration). **Passed after SME review.** |

| | | | | | | |
|---|---|---|---|---|---|---|
| **21** | 575 | 245 | Pass (0.27, 1.2%) | Pass (0.42, 3.1%) | Pass (1062.0N, 4.1%) | Pass |
| **22** | 338 | 245 | Pass (0.39, 1.7%) | Pass (0.47, 3.8%) | Pass (1249.4N, 6.9%) | Pass |
| **23** | 0 | 103 | Pass (0.39, 1.7%) | Pass (0.44, 3.4%) | Pass (859.3N, 2.3%) | Pass |
| **24** | 155 | 245 | Pass (0.32, 1.4%) | Pass (0.35, 2.2%) | Pass (869.5N, 2.3%) | Pass |
| **25** | 162 | 245 | Pass (0.25, 1.1%) | Pass (0.4, 2.8%) | Pass (1156N, 5.3%) | Pass |
| **26** | 171 | 245 | Pass (0.34, 1.5%) | Pass (0.34, 2.1%) | Pass (770.6N, 1.7%) | Pass |
| **27** | 437 | 245 | Pass (0.29, 1.3%) | Pass (0.38, 2.6%) | Fail (2182N, 54.0%) | Pass |
| **28** | 542 | 103 | Pass (0.75, 5.5%) | Fail (0.75, 13.1%) | Fail (2388.8N, 68.5%) | Fail 3/12 (UN & LN tension duration, UN Nij) |
| **29** | 260 | 136 | Pass (0.28, 1.2%) | Pass (0.38, 2.6%) | Pass (1221.8N, 6.3%) | Pass |
| **30** | 161 | 103 | Pass (0.28, 1.2%) | Pass (0.43, 3.2%) | Pass (518.8N, 0.83%) | Pass |
| **31** | 152 | 138 | Pass (0.34, 1.5%) | Pass (0.39, 2.7%) | Pass (1080.8N, 4.3%) | Pass |
| **32** | 225 | 103 | Pass (0.44, 2.1%) | Pass (0.4, 2.8%) | Fail (1601.7N, 17.3%) | Pass |
| **33** | 0 | 103 | Pass (0.54, 2.8%) | Fail (0.7, 10.7%) | Pass (1251.8N, 6.9%) | Fail 1/12 (LN compression duration). **Passed after SME review.** |

**Table 29** summarizes the instances of failure of each subcomponent of the MANIC for risk and injury specifications applied to the Data Set 1 tests. As the criteria becomes less conservative and the limits move from 5% risk of AIS 2+ to 5% risk of AIS 3+ and finally 10% risk of AIS 3+ the number of subcomponent failures is reduced. One or more components of the NIC failed 11 of the 33 test pre-SME review. Comparing the closest limits of the MANIC representative of the NIC (10% risk of AIS 3+) gives 18 failures of the MANIC compared to 11 of the NIC. For Data Set 1 evaluated at the 10% risk of AIS 3+ injury, the failure rate of the

small ATDs (less than 145 lbs, N = 20, mean airspeed=248 KEAS) was 70% (14 of 20) compared with 30.8% (4 of 13) for the large ATDs (greater than 245 lbs, N = 13, mean airspeed=261 KEAS).

**Table 29. Summary of MANIC Sub-Criteria Failures From Data Set 1**

| MANIC Failures | MANIC (Gx) | MANIC (Gy) | MANIC (Gz) | Total | MANIC Overall Test Failures | NIC Overall Test Failures |
|---|---|---|---|---|---|---|
| 5% AIS 2+ | 7 | 12 | 25 | 44 | 26/33 | 11/33 |
| 5% AIS 3+ | 3 | 11 | 23 | 37 | 24/33 | |
| 10% AIS 3+ | 0 | 5 | 17 | 22 | 18/33 | |

**Table 30**, **Table 31**, and **Table 32** provide the results of the MANIC when applied to another data set of full sequence ejection system testing. This data set tested fewer ejection seats and did not apply SME review. **Table 30** summarizes the performance of the 5% risk of AIS 2+ MANIC. Three of the 13 tests passed all elements of the 5% AIS 2+ MANIC compared with five of 13 passing all elements of the NIC. One or more of the 5% AIS 2+ MANIC elements failed on all eight of the tests that failed one or more elements of the NIC (eight of eight agreements). For all 13 tests the 5% AIS 2+ MANIC agreed with the NIC on 11 of them. There were two tests where the 5% AIS 2+ MANIC failed MANIC(Gz) where all elements of the NIC passed.

**Table 33** provides a detailed summary of the MANIC performance evaluating Data Set 2 at various AIS and injury risk levels. The 10% AIS 3+ MANIC results very closely matched the NIC results, failing nine of 13 tests compared to eight of 13. Additionally, the overall sensitivity of the MANIC at decreasing levels of injury risk and injury classification level behaved as expected. While matching the NIC results with the MANIC is not ultimately important, the fact that similar results were observed between the 10% AIS 3+ MANIC and the NIC provides some confidence that the MANIC is comparable to the legacy criteria. For Data Set 2 evaluated at the 10% risk of AIS 3+ injury, the failure rate of the small ATDs (103 lbs, N = 8, mean

airspeed=296 KEAS) was 63% (5 of 8) compared with 80% (4 of 5) for the large ATDs (245 lbs, N = 5, mean airspeed=322 KEAS), a different result than what was observed in Data Set 1. The higher failure rate of small ATDs than large ATDs in Data Set 1 and the higher failure rate of large ATDs than small ATDs in Data Set 2 could be due the fact that the large ATDs in Data Set 2 had the highest average ejection airspeed (322.2 KEAS) of the four groups (Data Set 1 small ATD, Data Set 1 large ATD, Data Set 2 small ATD, Data Set 2 large ATD). While the qualification test experimental set up is designed to test at the margins of the ejection envelope (i.e., smallest and largest ATDs at low and high ejection speeds) and not generate failure rate data for specific test conditions, it is interesting that small ATDs were not tested at airspeeds as high as large ATDs in Data Set 2.

**Table 30. Data Set 2 – AIS 2+ 5% MANIC Results Compared to NIC Results**

| | | | | **MANIC Results** | | **NIC Results** |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.56= 5% AIS2+)** | **Gy (<0.48= 5% AIS2+)** | **Gz (<922N= 5% AIS3+)** | |
| **1** | 0 | 103 | Pass (0.21, 1.2%) | Pass (0.31, 1.6%) | Pass (491.3N, 1.6%) | Pass |
| **2** | 0 | 103 | Pass (0.24, 1.4%) | Pass (0.31, 1.6%) | Pass (578.5N, 2.0%) | Pass |
| **3** | 174 | 245 | Pass (0.36, 2.2%) | Pass (0.31, 1.6%) | Fail (1535.1N, 21.7%) | Pass |
| **4** | 435 | 103 | Fail (1.28, 49.9%) | Fail (1.1, 80.7%) | Fail (3589.9N, 98.6%) | Fail 4/12 (UN & LN tension duration & Nij) |
| **5** | 435 | 103 | Pass (0.53, 4.3%) | Fail (0.52, 6.6%) | Fail (2029.1N, 51.4%) | Fail 1/12 (UN tension duration) |
| **6** | 426 | 103 | Pass (0.53, 4.3%) | Pass (0.47, 4.7%) | Fail (1860.3N, 40.1%) | Fail 1/12 (UN Nij) |
| **7** | 442 | 103 | Fail (0.72, 9.1%) | Fail (0.81, 35.2%) | Fail (3043.7N, 94.3%) | Fail 5/12 (UN & LN tension duration, LN shear duration, UN & LN Nij, UN MIx) |
| **8** | 601 | 245 | Pass (0.51, 4.1%) | Pass (0.45, 4.1%) | Fail (2570.5N, 82.1%) | Fail 1/12 (UN Nij) |

| | | | | | | |
|---|---|---|---|---|---|---|
| **9** | 434 | 103 | Pass (0.40, 2.6%) | Pass (0.38, 2.6%) | Fail (1278.1N, 12.1% | Fail 1/12 (UN tension duration) |
| **10** | 251 | 245 | Fail (0.72, 9.1%) | Fail (0.58, 9.7%) | Fail (3076.2N, 94.7%) | Fail 2/12 (UN & LN Nij) |
| **11** | 0 | 245 | Pass (0.15, 0.96%) | Pass (0.15, 0.52%) | Fail (1087.6N, 7.6%) | Pass |
| **12** | 197 | 103 | Fail (0.64, 6.7%) | Fail (0.69, 18.9%) | Fail (2376.6N, 73.1%) | Fail 6/12 (UN tension duration, UN Nij, UN MIx, LN shear duration, LN MIx, LN MIz) |
| **13** | 585 | 245 | Pass (0.52, 4.2%) | Pass (0.45, 4.1%) | Fail (2865.1N, 91.1%) | Pass |

**Table 31. Data Set 2 – AIS 3+ 5% MANIC Results Compared to NIC Results**

| | | | | **MANIC Results** | | **NIC Results** |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.72= 5% AIS3+)** | **Gy (<0.53= 5% AIS3+)** | **Gz (<1136N= 5% AIS3+)** | |
| **1** | 0 | 103 | Pass (0.21, 0.97%) | Pass (0.31, 1.9%) | Pass (491.3N, 0.77%) | Pass |
| **2** | 0 | 103 | Pass (0.24, 1.1%) | Pass (0.31, 1.9%) | Pass (578.5N, 0.99%) | Pass |
| **3** | 174 | 245 | Pass (0.36, 1.6%) | Pass (0.31, 1.9%) | Fail (1535.1N, 14.7%) | Pass |
| **4** | 435 | 103 | Fail (1.28, 25.0%) | Fail (1.1, 44.2%) | Fail (3589.9N, 98.7%) | Fail 4/12 (UN & LN tension duration & Nij) |
| **5** | 435 | 103 | Pass (0.53, 2.7%) | Pass (0.52, 4.8%) | Fail (2029.1N, 42.7%) | Fail 1/12 (UN tension duration) |
| **6** | 426 | 103 | Pass (0.53, 2.7%) | Pass (0.47, 3.9%) | Fail (1860.3N, 31.3%) | Fail 1/12 (UN Nij) |
| **7** | 442 | 103 | Fail (0.72, 5.0%) | Fail (0.81, 16.7%) | Fail (3043.7N, 93.8%) | Fail 5/12 (UN & LN tension duration, LN shear duration, UN & LN Nij, UN MIx) |
| **8** | 601 | 245 | Pass (0.51, 2.6%) | Pass (0.45, 3.5%) | Fail (2570.5N, 78.8%) | Fail 1/12 (UN Nij) |
| **9** | 434 | 103 | Pass (0.40, 1.8%) | Pass (0.38, 2.6%) | Fail (1278.1N, 7.4% | Fail 1/12 (UN tension duration) |

| | | | | | | |
|---|---|---|---|---|---|---|
| **10** | 251 | 245 | Fail (0.72, 5.0%) | Fail (0.58, 6.3%) | Fail (3076.2N, 94.4%) | Fail 2/12 (UN & LN Nij) |
| **11** | 0 | 245 | Pass (0.15, 0.8%) | Pass (0.15, 0.9%) | Pass (1087.6N, 4.4%) | Pass |
| **12** | 197 | 103 | Pass (0.64, 3.9%) | Fail (0.69, 10.2%) | Fail (2376.6N, 67.7%) | Fail 6/12 (UN tension duration, UN Nij, UN MIx, LN shear duration, LN MIx, LN MIz) |
| **13** | 585 | 245 | Pass (0.52, 2.7%) | Pass (0.45, 3.5%) | Fail (2865.1N, 89.9%) | Pass |

**Table 32. Data Set 2 – AIS 3+ 10% MANIC Results Compared to NIC Results**

| | | | **MANIC Results** | | | **NIC Results** |
|---|---|---|---|---|---|---|
| **Test** | **Speed (KEAS)** | **ATD Mass (lbs)** | **Gx (<0.95= 10% AIS3+)** | **Gy (<0.68= 10% AIS3+)** | **Gz (<1388N= 10% AIS3+)** | |
| **1** | 0 | 103 | Pass (0.21, 0.97%) | Pass (0.31, 1.9%) | Pass (491.3N, 0.77%) | Pass |
| **2** | 0 | 103 | Pass (0.24, 1.1%) | Pass (0.31, 1.9%) | Pass (578.5N, 0.99%) | Pass |
| **3** | 174 | 245 | Pass (0.36, 1.6%) | Pass (0.31, 1.9%) | Fail (1535.1N, 14.7%) | Pass |
| **4** | 435 | 103 | Fail (1.28, 25.0%) | Fail (1.1, 44.2%) | Fail (3589.9N, 98.7%) | Fail 4/12 (UN & LN tension duration & Nij) |
| **5** | 435 | 103 | Pass (0.53, 2.7%) | Pass (0.52, 4.8%) | Fail (2029.1N, 42.7%) | Fail 1/12 (UN tension duration) |
| **6** | 426 | 103 | Pass (0.53, 2.7%) | Pass (0.47, 3.9%) | Fail (1860.3N, 31.3%) | Fail 1/12 (UN Nij) |
| **7** | 442 | 103 | Pass (0.72, 5.0%) | Fail (0.81, 16.7%) | Fail (3043.7N, 93.8%) | Fail 5/12 (UN & LN tension duration, LN shear duration, UN & LN Nij, UN MIx) |
| **8** | 601 | 245 | Pass (0.51, 2.6%) | Pass (0.45, 3.5%) | Fail (2570.5N, 78.8%) | Fail 1/12 (UN Nij) |
| **9** | 434 | 103 | Pass (0.40, 1.8%) | Pass (0.38, 2.6%) | Pass (1278.1N, 7.4% | Fail 1/12 (UN tension duration) |
| **10** | 251 | 245 | Pass (0.72, 5.0%) | Pass (0.58, 6.3%) | Fail (3076.2N, 94.4%) | Fail 2/12 (UN & LN Nij) |

| | | | | | | |
|---|---|---|---|---|---|---|
| **11** | 0 | 245 | Pass (0.15, 0.8%) | Pass (0.15, 0.9%) | Pass (1087.6N, 4.4%) | Pass |
| **12** | 197 | 103 | Pass (0.64, 3.9%) | Fail (0.69, 10.2%) | Fail (2376.6N, 67.7%) | Fail 6/12 (UN tension duration, UN Nij, UN MIx, LN shear duration, LN MIx, LN MIz) |
| **13** | 585 | 245 | Pass (0.52, 2.7%) | Pass (0.45, 3.5%) | Fail (2865.1N, 89.9%) | Pass |

**Table 33. Summary of MANIC Sub-Criteria Failures From Data Set 2**

| **MANIC Failures** | **MANIC (Gx)** | **MANIC (Gy)** | **MANIC (Gz)** | **Total** | **MANIC Overall Test Failures** | **NIC Overall Test Failures** |
|---|---|---|---|---|---|---|
| **5% AIS 2+** | 4 | 5 | 11 | 20 | 11/13 | 8/13 |
| **5% AIS 3+** | 3 | 4 | 10 | 17 | 10/13 | |
| **10% AIS 3+** | 1 | 3 | 9 | 13 | 9/13 | |

These feasibility demonstration applications of the MANIC to real world data are a limited sensitivity analysis of the MANIC. The performance of each configuration of the pilot-scale MANIC in both Data Set 1 and Data Set 2 demonstrate the robust nature of the MANIC construct. The ability to customize the MANIC to specified levels of injury risk and injury classification is a useful tool in the acquisition and safety fields.

## Discussion

The purpose of this study was to outline methods to develop three axis-specific sub-criteria. When applied together, these criteria constitute pilot-scale multi-axial neck injury criteria to the aid development of vehicle safety systems to include military aircraft escape systems incorporating HMDs. The complete MANIC was applied to sample escape system data sets and compared to legacy criteria

Small PMHS data sets were a limiting factor in this study. For increased confidence in the MANIC(Gx) risk functions more PMHS experimental data is needed. -Gx data from ten or more additional subjects would increase statistical power and remove instances where the

difference between the AIS 2+ and 3+ risk functions depends on data from a single subject. New -Gx PMHS load and injury data should be collected under aviation-specific accelerative loading profiles following the methods used by the FAA to construct the side impact neck injury criterion (FAA, 2011). Incorporating new PMHS neck load and injury data with aviation specific accelerative injury loading is required to make the -Gx risk function more robust and applicable. The PMHS data used to construct the present MANIC(Gx) risk function had extreme AIS values, only one subject experienced neck injury other than AIS 0 or 6. Future human subject testing should also observe all six primary neck loads. If future -Gx experiments are able to provide the complete six load and AIS injury data it should be investigated if a full 6 load criteria structure would add to the injury prediction ability over and above the two-force combination of *Fz* and *My*. Frontal impact with its robust research history in the automotive environment is well understood, thus it is likely that the *Fz* and *My* loads adequately limit neck injury in this mode of acceleration.

The MANIC(Gy) risk function is the most robust of the three constructed in this study for a few reasons. It incorporates the most (five of the six) primary neck loads of any of the MANIC sub criteria. It also includes the most recent and arguably the most applicable PMHS experimental data since the experimental setup was designed for the purpose of evaluating subject response to aviation side impact. PMHS data for MANIC(Gx) incorporated relatively old, automotive experimental data. It would be ideal to have –Gx data from an aviation-focused, ejection-specific, experimental setup. The MANIC(Gz) incorporated multiple PMHS data sets, one of which was designed to replicate ejection-like tensile loading, though the other PMHS data sets were originally automotive research. Additionally, human subject experimental data which includes *Mx* is required for future, complete 6-load MANIC(Gy) development.

The tensile criteria presented in this work (-Gz) may be adequate as a criteria in that it incorporates a fairly comprehensive, robust, 12-subject PMHS study (Yliniemi et al., 2009) combined with other PMHS tensile studies and a sizeable human subject data set. It may be possible to augment the criteria with new human subject tolerance to pure tension. Since tensile forces have been shown a primary pathway for injury, incorporating a pure tension limit into the MANIC seems warranted (FAA, 2011).

The most significant lack of data preventing the development risk function development exists in the +Gz axis of acceleration. Data from a future +Gz PMHS experiment with a subject size of at least 10, is crucial for the future development of a MANIC(+Gz) risk function to capture neck injury risk from the catapult phase of ejection. As suggested previously for future –Gx experiments, future +Gz experiments should also follow the FAA side facing injury criterion construction protocol by observing neck load and injury at different levels of aviation-specific accelerations. Additionally human subject experiments which include all six primary neck loads are needed to ensure a complete MANIC(+Gz) could be developed in the future.

A few comments about the forces included and not included in the complete MANIC are warranted. It should be noted that in the complete criteria comprised of the risk functions from the three axes, *Mx* (coronal moment or side bending) was not included based upon the availability of existing data. In pure side facing accelerative input experiments the human head and neck kinematics observed did not include pure coronal moment, but a combination of twisting, coronal moment, and then flexion. The FAA tensile criteria for side facing aircraft seats does not include *Mx*, though it assumes some moment of unspecified magnitude is present along with neck tension as a necessary precondition for injury (FAA, 2011). While future improvements to the MANIC should include *Mx* if possible, the absence of this moment does not significantly detract from the value of the pilot MANIC. Additionally, *Fz* was included three

times each in MANIC(Gx), MANIC(Gy), and MANIC(Gz). *Fx* and *Fy* were included once each in MANIC(Gy). *My* was included twice in both the MANIC(Gx) and the MANIC(Gy). *Mz* was included once in MANIC(Gz). The instances of redundancy are acceptable because the axis specific risk curves for each individual axis provide protection for the primary loads contributing to injury experienced in that axis. Thus when applied to a dynamic accelerative environment like a full sequence ejection, the complete MANIC should limit peak loads for each direction of accelerative input.

A limitation of the MANIC is that no data from experiments where the primary loading was +Gz (representing the catapult phase of ejection and imparting compression to the cervical spine) was incorporated into the injury risk curves in this model. Compression (-Fz) is accounted for in the MANIC(Gx), but the Gx risk curve was constructed with frontal impact data where neck compression is not a primary load. While risk criteria exist to protect the thoracic and lumbar spine from injury during the catapult phased and few real world neck injuries are observed from the catapult phase mode of loading (Salzar et al., 2009; Lewis, 2006), this remains a limitation of the pilot neck injury criteria developed in this study. Another limitation of this study is the direct application of the human based MANIC to the Hybrid III aerospace ATD neck loads observed in the real world ejection data without a transfer function. Future research should explore the necessity and development of such a transfer function. This research could include collecting data including Hybrid III and human/cadaver neck loads under similar Gx, Gy, and Gz laboratory acceleration levels; then a mathematical relationship between the two data sets could be developed. The resulting function could be applied to ATD data collected during future system verification tests to estimate the likely human loads and applying the estimate in calculation of the MANIC.

## Conclusions

This paper presented the methods used in the preliminary, pilot-scale development of a set of improved, US Air Force multi-axial neck injury criteria which may be used as the foundation for full criteria development to be used to aid development and evaluation of escape systems incorporating HMDs. The MANIC meets the USAF escape system oversight office requirements for neck injury criteria. It is based upon human and PMHS data and incorporates survival analysis to generate axis specific risk functions. The MANIC was then applied to two sample escape system data sets and its performance was assessed compared to the NIC. The pilot MANIC demonstrates attributes that provide potential value over the legacy NIC. The method used to construct the MANIC risk functions allow for the user to limit varying levels of specified AIS injury at various risk percentages. When applied to system testing, the MANIC provides specific risk quantification for system safety evaluation and enables decision-makers to incorporate the probability of risk into tradeoff decisions. The ability to quantify the risk posed by an HMD or escape system at varying injury levels and varying risk levels is an important feature of the MANIC that is not present in legacy criteria.

# VIII. Incorporating Air Force Ejection Neck Injury Criteria into DoD Architecture

## Introduction

Ensuring pilot safety is of utmost importance within the Air Force (AF). Throughout the design, production, testing, and operation of manned flight systems, great expense and care is taken to ensure that these systems do not subject the pilot to an unacceptable risk of injury. The goal of this chapter is to incorporate the Department of Defense Architecture Framework (DoDAF) to depict the architecture of the systems, functions, and data involved in designing helmet mounted display (HMD) systems while incorporating neck injury criteria (DoDAF, 2009). Multiple organizations, actors, interfaces, data collection, and flows are involved in developing and implementing such criteria. Stakeholders in this system of systems include the AF acquisitions community, Air Force Research Laboratory (AFRL) biomechanics testing branch, the research community, the operational flying community, defense contractors, the 846th Test Squadron at Holloman AFB, which performs developmental testing of escape systems, and the AF flight safety community. The DoDAF viewpoints developed in this chapter provide insight into the system and illuminate potential interface issues as well as highlight the breadth of organizations involved in developing, implementing, testing, and complying with the requirements of escape system neck injury criteria.

## Viewpoints

### Operational Concept

The purpose of an OV-1 is to provide a graphical depiction of the operational concept. Figure 39 illustrates the OV-1 viewpoint, the high level operational concept graphic. It depicts the major functions involved in developing and implementing pilot neck injury criteria.

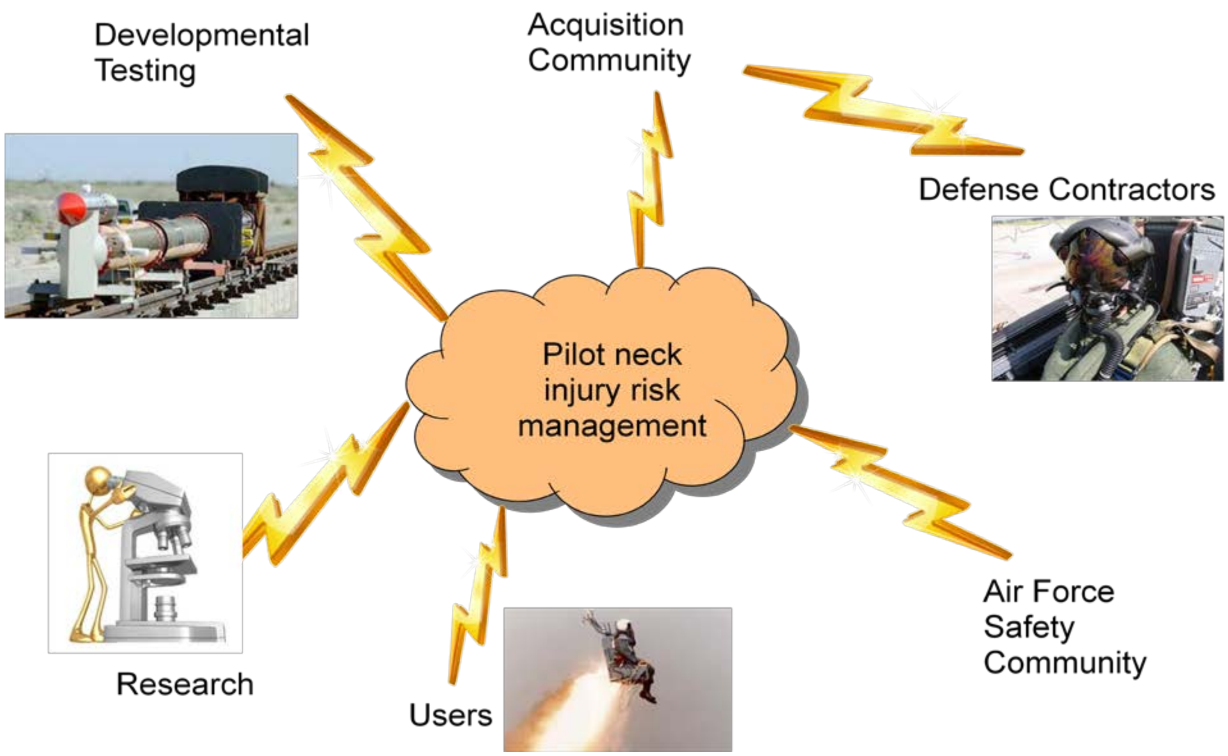


**Figure 39. High Level Operational Concept Graphic (OV-1)**

Details for each element in this high level architectural view are provided in the next view, the operational resource flow description.

### Operational Resource Flow Description

The next viewpoint to consider is the Operational Resource Flow Description, the OV-2. The goal of this view is to describe a "should-be" viewpoint of the systems involved and the resource flows between operational activities in the collection, processing, and storage of data to support neck injury criteria. The OV-2 is shown in **Figure 40**. The four major organizations or systems involved in this view include the testing unit, the system development and acquisition offices, the aircraft flying units, and the injury risk research institutions. The operational activity

descriptions summarize the general functions of each part of the system, and the arrows between them depict in general the types of resource flows exchanged between each part of the system.

The testing unit is the Holloman AFB test squadron (or equivalent contractor testing unit, e.g., Martin Baker's high speed test track at Langford Lodge, Ireland), which performs the developmental testing on new escape systems and modifications to existing escape systems. The test squadron performs ejection tests of the systems on the rocket sled track at various representative airspeeds with instrumented anthropometric test devices (ATDs) and records the performance of the entire system, to include the neck loads experienced by the ATD during the ejection sequence. If an HMD is part of the equipment required to operate the aircraft, then it is included in testing as part of the escape system being evaluated for safety. The result of this series of tests is evaluated by the application of neck injury criteria, and it is determined if the system passes or fails the developmental testing. If the neck injury criteria are exceeded, the system does not pass developmental testing and the exceedence must be addressed by the contractor. This is the point in the process where the set of neck injury criteria are implemented and used as a critical decision making tool. The set of criteria is what ensures an acceptable level of safety is built into the escape system and is the primary assessment tool to determine that a pilot's neck will not be put at undue risk as a result of the operation of the escape system. Operational testing of escape systems is not accomplished due to a shortage of motivated volunteer test pilots.

The system development and acquisition offices include the escape engineering community that provides program managers with a common set of overarching requirements and standards to be implemented in all escape system development. They include the program offices themselves who manage cost, schedule, and performance of weapon systems. Specific to the topic at hand, they ensure escape systems are designed and produced to meet requirements,

which includes meeting acceptable neck injury risk standards.  Note that it was decided in the construct of this viewpoint to include the defense contractors who manufacture the systems within the acquisition function, though in reality they are separate entities.  The fact that the government acquisition program offices and defense contractors work together to accomplish the same function of developing and manufacturing the system lends itself to treat them as one from an architectural standpoint, though it is understood they have different roles in the system acquisition process.  The system development and acquisition offices provide test plans, test articles, and evaluation criteria to the testing node.  Additionally, they provide injury criteria requirements and guidance to the injury risk research institutions, which direct the scientific and engineering community's research focus.

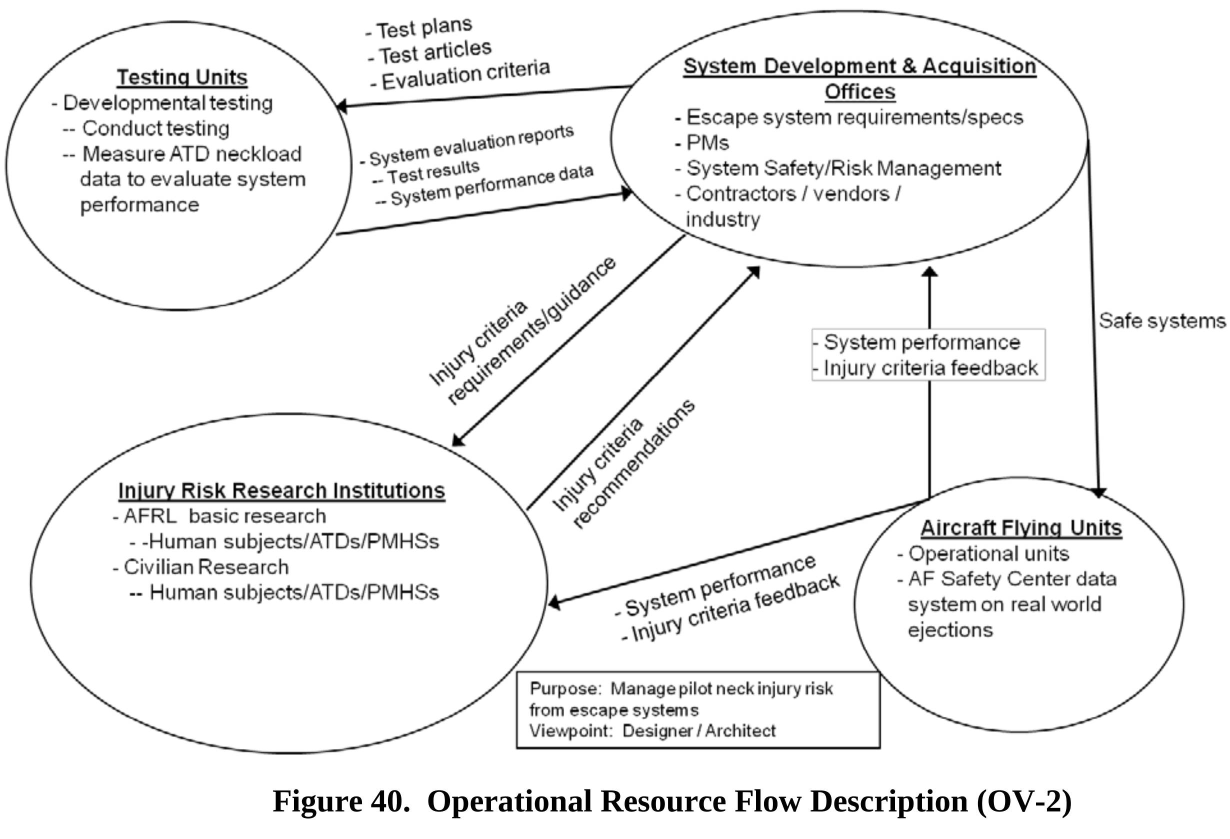


**Figure 40.  Operational Resource Flow Description (OV-2)**

The aircraft flying units are comprised of the units operating the weapon system and the AF Safety Center. The system development and acquisition function delivers the system to the operational units who are the end users. Conversely, the aircraft flying units provide system performance feedback to the system development and acquisition function. The AF Safety Center investigates and catalogues available information from the investigations they perform on all real world aircraft ejection events. This database is used by the injury risk research function to assess and validate injury criteria with real world data.

The final entity in the OV-2 is the injury risk research institutions, which include the human subject, ATD, and post mortem human subject (PMHS) research performed by AFRL and civilian institutions for application to escape system safety. This research covers a variety of topic areas but generally focuses on evaluating new escape technologies, HMD systems, and furthering understanding of the biomechanical affects of ejection forces on the human body. These institutions develop injury criteria through research and provide recommendations to the system development and acquisition offices based on this engineering judgment.

### Activity Decomposition Tree

The final viewpoint considered is the operational activity decomposition tree, the OV-5a (Figure 41). This view helps to identify the activities required to manage pilot neck injury risk due to escape systems, which improved ejection neck injury criteria will greatly support. Any lapse or interruption to any of these critical activities will impact the development and implementation of improved neck injury criteria.

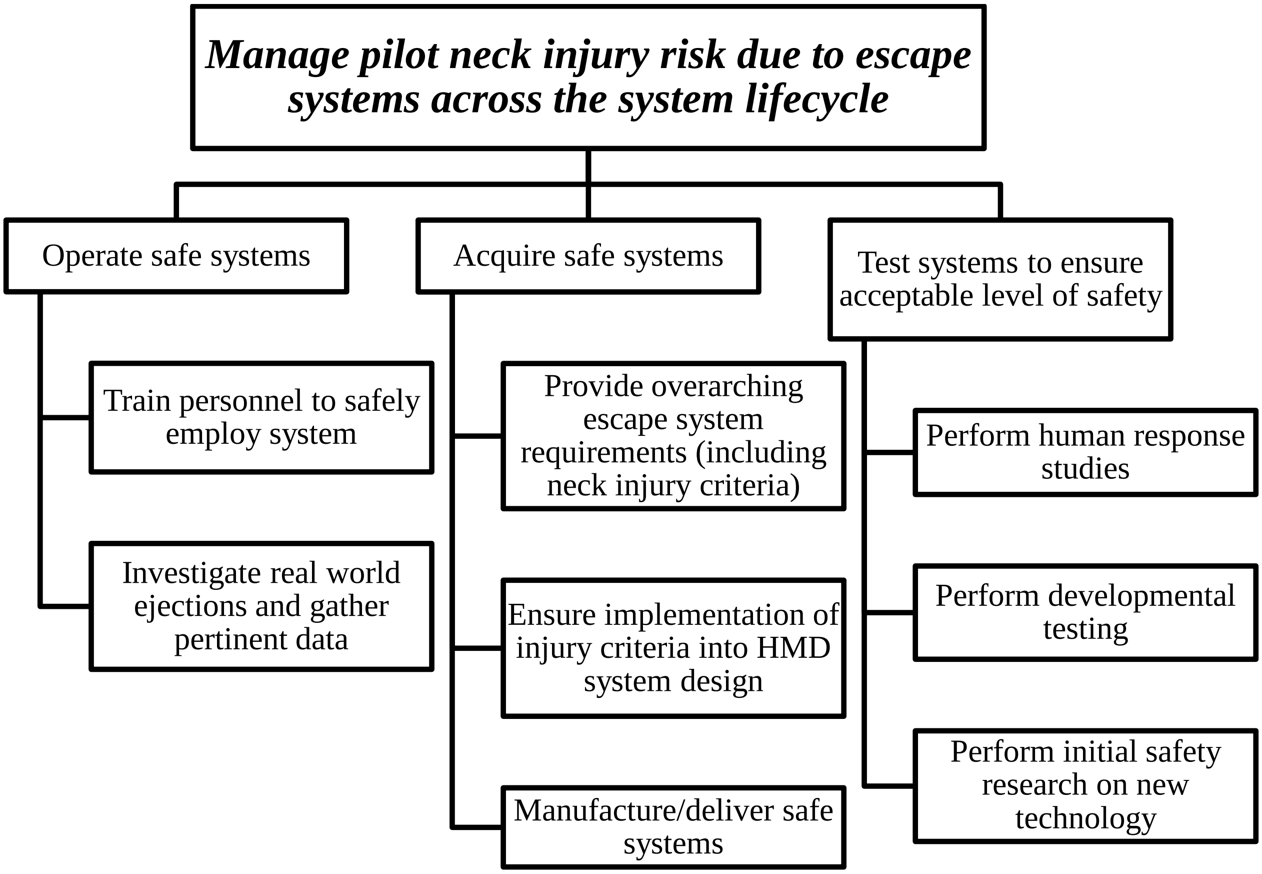


**Figure 41. Operational Activity Decomposition Tree for Management of Ejection Neck Injury (OV-5a)**

### Significant Design Considerations

Throughout the process of constructing the various DoDAF viewpoints, a few key issues and design considerations arose. First, from the OV-2 it is apparent that the data flow from the injury risk research institutions to the system development and acquisition offices is crucially important to developing and implementing improved neck injury criteria. If this flow shuts off, the acquisition community has no technical scientific or engineering foundation from which to establish and promulgate evidence-based neck injury criteria that meet its requirements. This

issue has been observed in the real world system. A lack of interest and funding over the past decade on the scientific research surrounding manned flight versus unmanned flight research by AFRL has significantly slowed research and data collection necessary for the development and implementation of improved neck injury criteria. The result is outdated injury criteria still in use in the Escape Systems Joint Service Specifications Guide, the primary requirements document put forth by the AF Life Cycle Management Center (AFLCMC) engineering home office to guide the development of all escape systems throughout the program offices that fall under the AFLCMC (Department of Defense Joint Service Specification Guide; Crew Systems Emergency Egress Handbook, 1998). Recently, resurgent funding and attention has been given to human performance and biomechanics research, and the research data flow to the acquisition organizations is increasing. The importance of the flow of information from the research institutions to the system development and acquisition offices in the OV-2 became very noticeable during the development of these viewpoints.

The second consideration that arose during the application of these architecture viewpoints to the AF ejection neck injury criteria environment was the importance of operational data to validate the improved neck injury criteria and the obvious lack of that data flow under the current system. This need could possibly be remedied by instrumenting a small sample of pilots with accelerometers to acquire real world neck load data. If even a few actual ejections were captured, the criteria validation and resultant system design benefits would be worth the cost and effort. Even if no ejection data were captured, the neck load data from the sample of instrumented pilots would provide a baseline of neck loads experience by fighter pilots over the duration of a mission. This would provide much needed real-world data to fuel initial studies on the impact of pilot fatigue due to additional head supported mass.

Third, the significant importance of the escape systems guidance office within the AFLCMC became clear. Vital functions performed by this office include providing neck injury criteria requirements to the injury risk research node as well as providing the overarching standards and requirements for escape systems to the program offices. The appropriate funding for resources and personnel should be provided by AFLCMC leadership to ensure this function can perform at its maximum potential. Otherwise, all programs are likely to suffer for lack of clear and current escape injury criteria. The cost of this inattention is likely to manifest in the form of fielding unsafe escape systems.

Finally, this system architecture highlighted the diverse stakeholders involved in developing and implementing improved neck injury criteria. This will be helpful in future communications with these stakeholders to highlight the important interfaces and critical data flows that are required amongst the stakeholders for successful management of AF pilot neck injury risk in the future as escape and HMD systems are developed, tested, and fielded.

# IX. A Human Systems Integration Analysis of Helmet Mounted Displays

## Chapter Overview

The paper that comprises this chapter was accepted for publication in the SAFE Association Journal. This paper outlines the HMD system design trade space and proposes a preliminary model to maximize the ratio of total system performance and total ownership cost (TOC). In this paper the application of the neck injury risk criteria to quantify risk associated with head supported mass is demonstrated. Additionally, the -Gx criteria developed in Chapter IV is applied to establish the link between improved neck injury criteria and HSI.

## Abstract

Helmet mounted displays (HMDs) provide increased capability to advanced aircraft systems but also add mass to the pilot's head. This mass potentially increases fatigue, degrades pilot scan patterns, and potentially increases chronic, as well as acute injury during accelerative loading. From a Human Systems Integration (HSI) perspective, HMD capabilities should be selected to maximize performance and minimize system total ownership costs (TOC). Unfortunately, a clear method does not exist for performing this HSI tradeoff analysis to include safety (acute neck injury), occupational health (chronic neck injury), human factors engineering (performance and fatigue), and survivability. This study utilized content analysis and data to propose a qualitative model of the impacts HMDs have on HSI. Further, recent research on neck injury risk criteria was applied to quantify the impacts of helmet mass on the ejection safety portion of the model. A methodology for the formulation of a quantitative model of parameters influencing the HSI impacts of HMDs was developed. This study illustrates the difficulty in formulating a rigorous optimization of HSI parameters for a HMD. If quantitative HSI assessments of realistic system performance and TOC are to be conducted, additional research will be required.

## Introduction

Human Systems Integration (HSI) is a method for addressing human-centered concerns during system design. Department of Defense Instruction (DoDI) 5000.02 directs that HSI is to be applied to "optimize total system performance, minimize total ownership cost" and ensure that the system accommodates the user population (DoD, 2008). However, "optimizing total system performance" may not be consistent with "minimizing Total Ownership Costs" (TOC). As a result, this criterion might be reinterpreted to maximizing the ratio of total system

performance to TOC. Other possible optimization formulations consist of either maximizing total system performance subject to some maximum TOC constraint, or minimizing TOC subject to a total system performance minimum constraint. Regardless of this interpretation, it is first necessary to quantify each of these attributes to enable “optimization” (e.g., minimization or maximization).

HSI has gained emphasis, both within military acquisition (Booher, 2003) and the systems engineering community (Madni, 2009). The HSI concept assumes that by associating human-centered concerns with human-centered domains, one can arrive at an improved system solution. This solution considers the impact of these concerns within each domain and synthesizes the results to understand the impact of potential system trades on total system performance and TOC. These domains, which often include manpower, personnel, training, human factors, occupational health, and safety, represent areas of human-centered technical expertise which ideally can assess the impact of system trades on performance and cost (AF HSI Office, 2009). This paper addresses the question of whether optimization can be achieved as instructed by DoDI 5000.02 for human-centered systems and proposes a methodology for the formulation of a quantitative model of parameters to perform HSI trade space analysis on HMD systems.

To address this question, an analysis is applied to the design of helmet mounted display (HMD) systems for fixed-wing aircraft. Within the human factors literature, it has been documented that the use of HMDs can improve pilot situation awareness (SA) as critical information can be displayed to the user without requiring visual search or fixation on head-down displays (Geiselman and Havig, 2011). However, as with many technologies the intended technical improvements often have negative unintended consequences. For example, such a display may increase the head supported mass to values beyond 5 lbs. Such an increase in head

supported mass for the entire mission duration is a significant departure from the operational procedures of legacy systems, and the totality of costs and performance are not well understood.

HMDs were selected for this analysis as they are used in numerous legacy fixed and rotary wing aircraft and are likely to be common human-machine interface equipment in the future of manned flight. Currently HMDs are in use on multiple Department of Defense (DoD) weapon systems (e.g., A-10, AH-64, F-15, F-16, F-18, C-130) and planned for the F-35 Joint Strike Fighter. In fact, this human interface technology is becoming more prevalent with increasing capability (VSI, 2013). Throughout the remainder of this paper it will be assumed that this trade space model for evaluation of HMD systems could be applied to a range of military applications including fixed wing, rotary wing, and mounted or dismounted ground operations.

The benefits of addressing HSI domains early in the systems acquisition lifecycle have been documented (Booher, 2003; INCOSE, 2011); however, these documented examples rely heavily on expert opinion rather than rigorous quantitative trade analysis. Limited quantitative HSI research exists in the literature, though Hardman has put forth quantitative HSI engineering methodologies in the areas of aircraft mishap prevention requirements and user interface design (Hardman, 2009). Numerous human factors studies have been performed which attempt to explain and quantify human visual and cognitive performance relative to military applications of HMDs (Rash et al., 2009), but this research has not been adequately translated into an overarching quantitative HSI application. This example illustrates the difficulties which arise when attempting to provide a quantitative HSI analysis within a practical (although constrained) systems engineering process. It should be noted that there are a number of system level trades involved with HMDs, but the purpose of this paper is to focus on those that are HSI oriented. Thus, items such as maintainability costs, sustainment costs, ruggedization, etc. will not be

included in the current analysis. That is, it is assumed that these items are fixed in the analysis. The purpose of this paper is to develop and present a methodology for the formulation of a quantitative model of parameters influencing the HSI impacts of HMDs.

### Applicable Definitions

As HSI is a multidisciplinary field, it is important to specifically define important terminology. First, monetary cost terms to clearly differentiate include Lifecycle Cost (LCC) and Total Ownership Cost (TOC). Both the DoD and the International Council on Systems Engineering (INCOSE) define LCC as the totality of acquisition and ownership costs of a system over its entire life to include concept, development, production, operation, sustainment and disposal (INCOSE, 2011; DAU, 2013a). LCC also includes indirect costs that can be reasonably linked to the system. TOC incorporates LCC, but also includes "related infrastructure or business processes costs not necessarily attributed to the program in the context of the defense acquisition system" to include medical care, which is especially germane to the current study. INCOSE provides less distinction, but perhaps allows for more flexibility, between its definition of LCC and TOC, incorporating into TOC some of the costs which the DoD considers in its definition of LCC (Rash et al., 2009). In general, however, INCOSE considers much of the HSI related costs in its definition of TOC, including personnel costs, training costs, costs of mishaps, and disability compensation and liability claims (Rash et al., 2009). What is important for the purposes of this paper is to establish that we aim to minimize TOC, which includes LCC plus medical care costs associated with the HMD system that may extend beyond the lifecycle of the program.

Although the research and development components of LCC are closely monitored during the acquisition lifecycle and unexpected expenses incurred during this phase of a system's lifecycle draw significant public scrutiny and compensatory legislation (WSARA, 2009), the operation and support components of LCC have drawn less attention. However, these costs have recently begun to draw similar scrutiny, although projections of these costs are more difficult (Ryan, 2012). Total ownership costs have received less focus than either acquisition or lifecycle costs as these costs can arise from unpredictable sources such as environmental contamination by an unknown carcinogen or other human systems hazards with consequences that are unknown or difficult to project.

While monetary costs are an important consideration within the current paper, it is also understood that system attributes intended to improve the performance of the operator or the system can additionally reduce the performance of one or more of these entities and therefore cost can also refer to loss of performance. Total System Performance refers to the quantifiable mission capability performance of the system to which the HMD contributes. In this study the aspects of total system performance not related to the HMD will be assumed to be constant and will not be considered since it is desired to analyze only the performance contributions of the HMD system. Operator Performance (in Figure 42 below) is specifically the cognitive, sensory, and physical human performance that is either enhanced or degraded by the HMD.

## HSI Analysis

### Outlining the Trade Space

A causal loop diagram was created, as shown in Figure 42, to depict a portion of the relationships affecting HMD utility from an HSI point of view. The traditional Systems

Engineering (SE) top down functional decomposition begins with capabilities (operational requirements) from which system requirements are generated. System functions are generated from system requirements and then allocated to individual components which are described by parameters. In Figure 42, the “HMD System Parameters” block represents those functions, and their descriptive design parameters, which were allocated to the HMD.

As shown in Figure 42, as the values of HMD system parameters increase, the performance of the operator should be expected to increase. Consider such scenarios as greater field of view, larger displays with more resolution, or additional night-vision functionality. In the diagram, a plus sign signifies that a change in the first entity causes a change in the same direction in the second entity, while a minus sign signifies that a change in the first entity results in a change in the opposite direction of the second entity. Improvements in operator performance would be expected to improve overall system performance. Improvements in operator performance will also likely reduce the probability that the pilot will need to eject from the aircraft, improving system survivability, which, in turn, may reduce the likelihood of acute neck injury during ejection from the aircraft.

This proposed increase in HMD system parameters often requires modification of HMD system hardware, which can increase the mass of the HMD. This mass is supported by and adds load to the human operator’s neck and spine. This increase in mass then tends to increase the fatigue of the operator, which is often considered within the Human Factors Engineering domain. Further, the likelihood of chronic neck injury occurring due to the repeated exposure of the neck to greater than natural forces as the pilot is exposed to accelerative environments, such as those posed by vibration, buffeting, or high rate maneuvers also increases. Chronic neck injury is often considered within the domain of occupational health. Finally, as the mass of the HMD increases, the likelihood and severity of acute neck injury also increases in the absence of

mitigating technologies (such as the Neck Protection Device on the JSF ejection seat), as the pilot may be exposed to high accelerations, for example during ejection. This effect of mass impacts the Safety domain. Note that each of these factors has the potential to decrease operator performance.

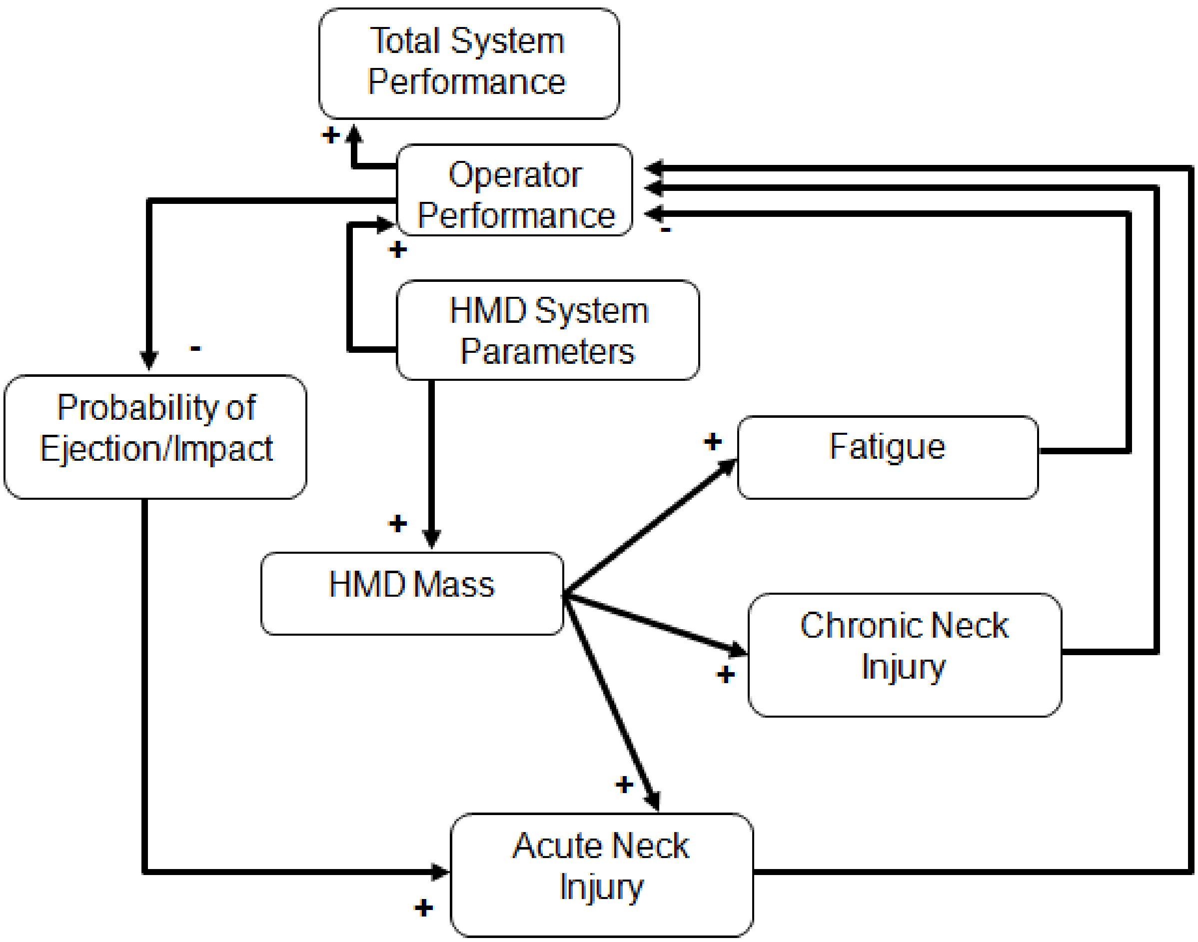


**Figure 42. Partial HMD HSI Causal Loop Diagram**

Fatigue directly impacts the performance of the operator throughout the mission, while chronic neck injuries and safety issues may reduce the time that an operator can perform within the platform, thereby increasing the costs for operator recruiting and training while reducing average operator experience. Therefore, the platforms might, on average, have less experienced operators who may have a lower performance than more experienced operators. Increases in operator performance are expected to increase total system performance. While it is understood that total system performance involves more than just the contribution of the operator's performance which is enhanced through HMD system parameters, the current analysis is limited only to this portion of total system performance.

During technology development, effort may be spent to reduce the effect of increases in HMD system parameters on HMD mass through the use of lighter materials, increases in technology integration, or other technological innovations. However, technology development requires investment, increasing development costs, which may increase or decrease TOC, as depicted in Figure 43. It is possible that the investment in HMD developmental costs during acquisition may decrease HMD mass, which would decrease the fatigue and chronic/acute neck injury costs associated with the additional mass. Assuming added HMD system parameters increase HMD mass, TOC is likely to increase as acute and chronic neck injuries increase. Decreases in the probability of ejection will likely decrease TOC. Therefore, it is intuitive that tradeoffs exist within the HMD design which influence total system performance and TOC.

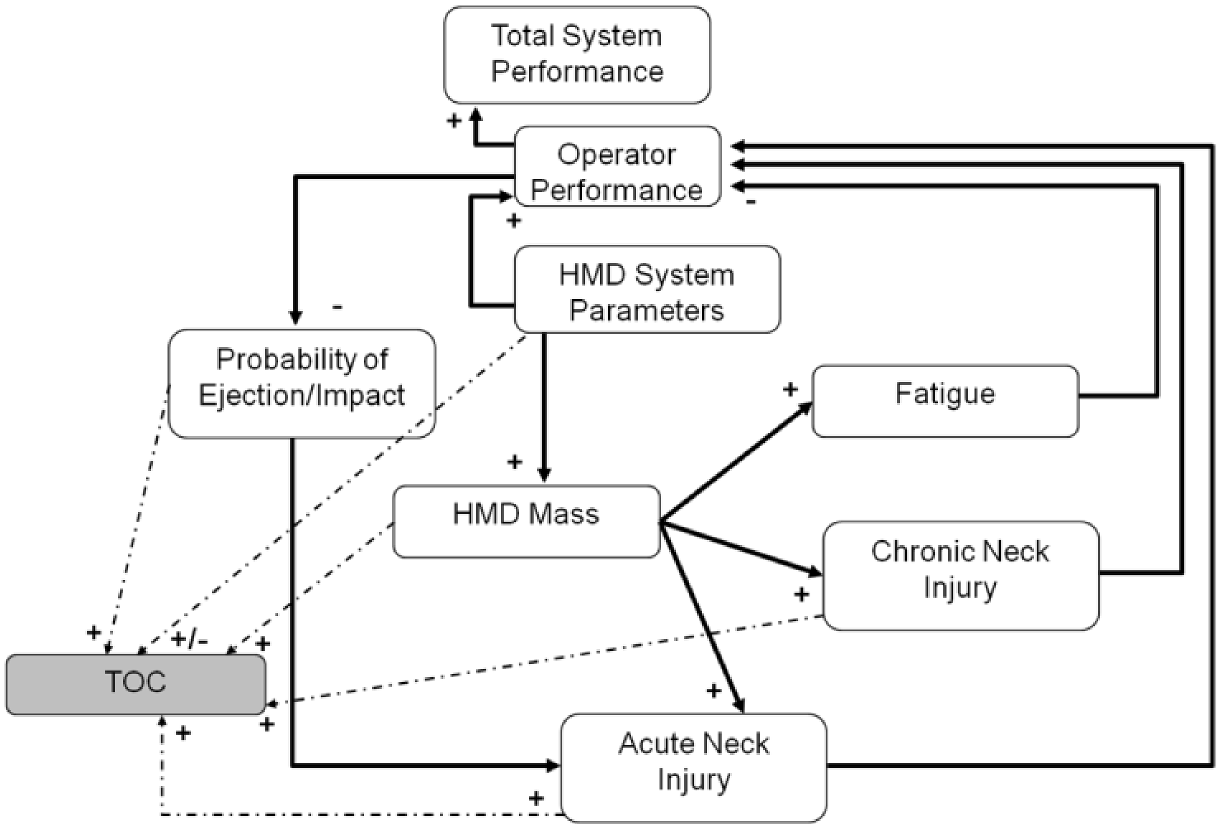


**Figure 43. TOC in Context of the Partial HMD Causal Loop Diagram.**

Although Figure 43 reflects a number of high level trades with respect to HMDs, it does not reflect any specific changes in HMD system parameters or the impact of these parameters on performance. As depicted in Figure 44, this added capability could, for example, include changes to HMD parameters. For instance design parameters of the HMD could include field of view (FOV), night vision, resolution, and binocular or monocular viewing. Each of these parameters may affect human performance and, therefore, total system performance. However, these relationships are often not well specified. Instead, the human factors literature will often associate changes in these HMD parameters with intermediate attributes, such as the ability of an operator to search for and detect a target, target the enemy, detect or understand platform motion, maintain comfort without eye strain, or determine the orientation of their aircraft, as shown in

Figure 44. While performance on these tasks is likely to influence operator performance and therefore total system performance, this relationship is often difficult to ascertain.

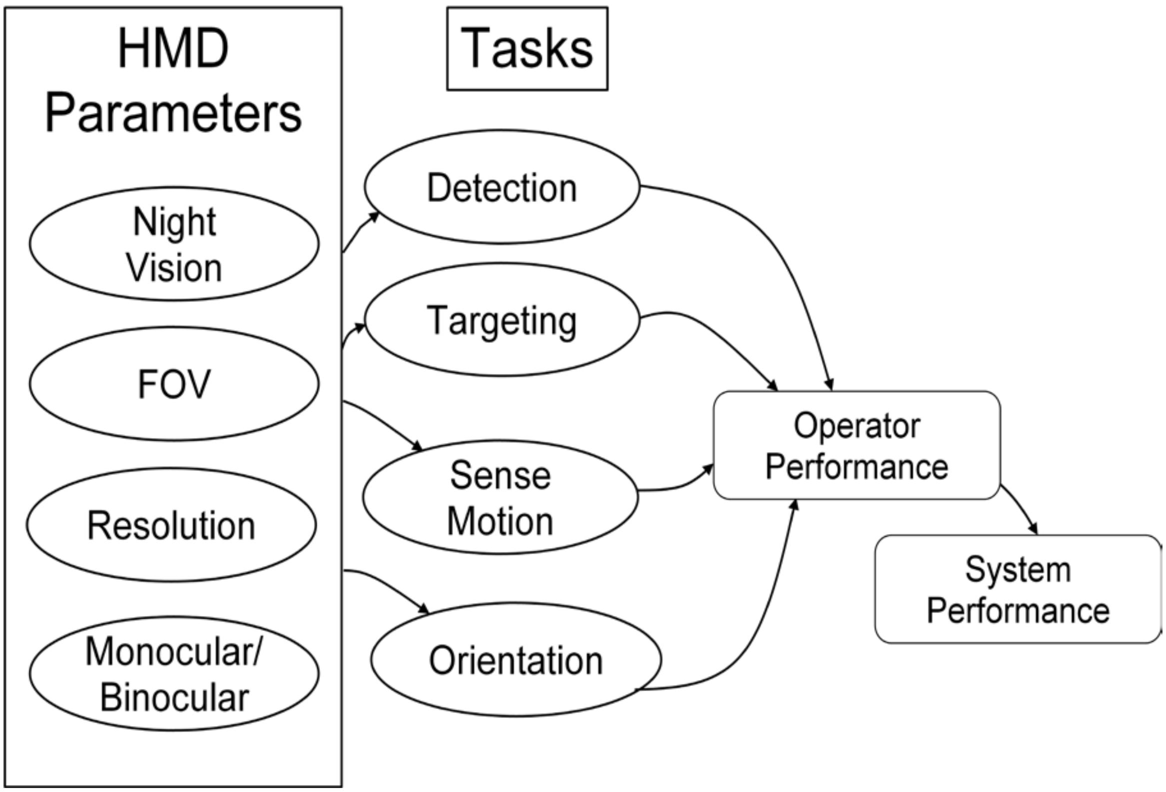


**Figure 44. HMD Parameters and Tasks That Contribute to System Performance**

In an analysis of the system, it is assumed that increases in HMD system parameters, such as the night vision and binocular optics are being added to increase operator performance and as a result, improve total system performance. Within the context and scope of HSI, however, increased operator performance comes at a cost, not only in monetary terms, but potentially in total system performance as indicated in Figure 43. For example, the cost is an

increase in HMD mass, which is likely to increase fatigue over mission length durations, decreasing operator performance. These unintended costs and negative impacts on operator performance must also be fully understood to quantify HSI tradeoffs. Both the benefits and the costs to human performance, safety, and health must be identified at the earliest point possible in system development to make the appropriate HSI cost and benefit trade decisions in the system design.

## Human Performance: Overview of Applicable HSI Domains

The benefit from adding HMD capability lies primarily in the HSI domains of human factors engineering and survivability. In the human factors engineering domain, benefits include increased SA, increased target cueing capability, and increased precision navigation capability. In the survivability domain, increased performance is realized from the previously mentioned human factors benefits (SA, target cueing, and precision navigation) each contributing to human performance, which increases survivability of the weapon system, thus increasing total system performance.

Incorporating HMD system parameters usually (though not always) requires adding mass to the unit, which increases the operator's head supported mass. While it may be argued that some added parameters might result in a decrease in head supported mass, for the purposes of this first order model it will be assumed that adding HMD system parameters results in increased head supported mass. The cost to operator performance of this additional head supported mass comes in the HSI domains of survivability and human factors. In the human factors domain, performance degradation comes in the form of neck fatigue caused by HMD mass, which potentially impacts the mental performance of the operator.

### ***Human Factors Engineering***

The AF HSI Handbook defines the human factors engineering domain of HSI as follows: "The comprehensive integration of human capabilities and limitations (cognitive, physical, sensory, and team dynamic) into systems design, to optimize human interfaces to facilitate human performance in training operation, maintenance, support and sustainment of a system (AF HSI Office, 2009)." This section analyzes the applicable beneficial components of human factors engineering applicable to HMD system parameters.

*Situation Awareness.* According to Rash et al., the chief objective of HMD designers is to maximize SA for the operator (Rash et al., 2009). Endsley has put forth a widely accepted three level definition of SA as "Level 1) the perception of the elements in the environment within a volume of time and space, Level 2) the comprehension of their meaning, and Level 3) the projection of their status in the near future (Endsley, 1995)." An Air Force definition of SA has been proposed by Geiselman as "A pilot's continuous perception of self and aircraft in relation to the dynamics of flight, threats, and mission, and the capability to forecast, then execute tasks based on the perception (Geiselman and Osgood, 1994)." Many technologies can be added to a HMD which would provide increased SA. Within a HMD, space for important symbology for system operation and user SA is at a premium. Increased resolution and FOV would help alleviate this problem. Geiselman has suggested that if additional information (specifically ownship status symbology) could be presented, it could add "operational utility of the HMD by increasing lethality and survivability for day, night, and all weather application (Geiselman, 2013)." While more information is not always better, it will be assumed in this analysis that the presentation will be designed in accordance with established human factors practices so as not to confuse or overwhelm the pilot's ability to obtain the necessary information from the display.

*Target Cueing.* The improved performance of HMDs gives tactical fighter aircraft a distinct advantage in targeting. According to Rash et al., “HMDs are ‘must have’ equipment on GEN-4 fighter aircraft, since high off-boresight weapons and visual cueing outweigh any aircraft-performance advantage during a dogfight (Rash et al., 2009).” A pilot’s ability to look and target with the HMD instead of with the nose of the aircraft, subjecting him/herself and the airplane to high G loading dramatically altered fighter pilot tactics, significantly increasing operator performance and total system performance. This same technology is incorporated into rotary wing HMDs for target cueing.

*Navigation.* An improved HMD with increased FOV, resolution, night vision and binocular capability would increase precision navigation performance. The flight information required for navigation could be better displayed and would allow the operator to better fuse navigation inputs thus improving this portion of the mental workload required during flight. HMDs allow the user to monitor important data without switching their visual attention from the operational environment to view the instrument panel, and then integrating information from the two disparate sources. Overall operator performance improves when key flight information is presented within the pilot’s line of sight (Rash et al., 2009). Pilots are able to detect changes within their field of view since the HMD allows them to keep their gaze forward (Rash et al., 2009). A well designed layout of the navigation information within the display area will enhance human performance in this area. Additionally, night vision would enable this same capability to be leveraged at night.

### *Survivability*

The AF HSI handbook defines the survivability domain of HSI as “The ability of a system, including its operators, maintainers and sustainers to withstand the risk of damage, injury, loss of mission capability or destruction. Survivability includes the elements of susceptibility, vulnerability, recoverability, and suitability (AF HSI Office, 2009).”

Many of the capability enhancements discussed previously in the human factors section, including SA, target cueing, and precision navigation also contribute to increased survivability as increases in operator performance will likely reduce the probability of platform loss. Increased SA is likely to reduce human error which could result in controlled flight into terrain, runway incursions, or mid-air collisions. For ground operators, increased SA prevents fratricide and provides increased overall battle space awareness, potentially preventing the enemy from becoming a destructive threat. Improved target cueing counters the adversary, improving blue force survivability. Precision navigation enhances maneuverability in low level terrain, specifically in rotary wing and tactical airlift operations, decreasing platform visibility.

## Example Trade Space Analysis: Human Performance

Development of a preliminary model begins with identifying quantifiable performance trade space. In this section, the influence of an example HMD system function on operator performance is explored, and a notional or approximate relationship is shown. For a fully developed model, the user could follow this methodology for the specific HMD system parameters of interest for their specific HMD trade space analysis.

The HMD components which add mass as well as influence operator performance ($p$) are described in the formulation below by the aggregation of performance-increasing parameters ($\bar{x}$)

(e.g. increased field of view, increased resolution, night vision, binocular system versus a monocular system, medical monitoring, laser eye protection, fusion, tracking accuracy, eye tracking, optics quality, on-head computing, impact/penetration protection, high speed visors for day and night configurations, transitions from day-to-dusk or night-to-dawn, etc.) and performance-degrading parameters ( $\bar{y}$ ) (e.g. HMD mass, extreme HMD center of gravity (CG)). This aggregation can be stated as seen in Equation 21.

$$p_{tot} = f[p(\bar{x})] - g[p(\bar{y})] \tag{21}$$

where $\bar{x} = \{x_1, x_2, ..., x_n\}$ and $\bar{y} = \{y_1, y_2, ..., y_n\}$

### *Example Human Performance Benefit: Field of View*

As an example of how one performance-increasing parameter ($x_n$) is quantified, a study linking HMD display FOV was analyzed. An element of SA is target detection. Nelson et al. explored the effects of FOV on operator performance, specifically detection of an oncoming aircraft (Nelson et al., 1998). The results are shown in **Table 34**. Operator target detection performance was observed to increase as a function of FOV.

**Table 34. Target Detection as a Function of Field of View**

| Field of View | Correct Detection (%) | Detection Distance (m) |
|---|---|---|
| 60x40 | 83 | 1800 |
| 150x70 | 91 | 2150 |

This data can be used to derive the model depicted in Figure 45. Although this model approximates the impact of FOV on human target detection, the relationship between this function and total system performance would require further study. Using a similar

methodology, research from other human performance studies could similarly define the link shown in Figure 44 between HMD parameters, operator tasks, and operator performance. Studies like the Nelson et al. research provide a quantifiable link for use in an overall HMD trade space analysis (Nelson et al., 1998).

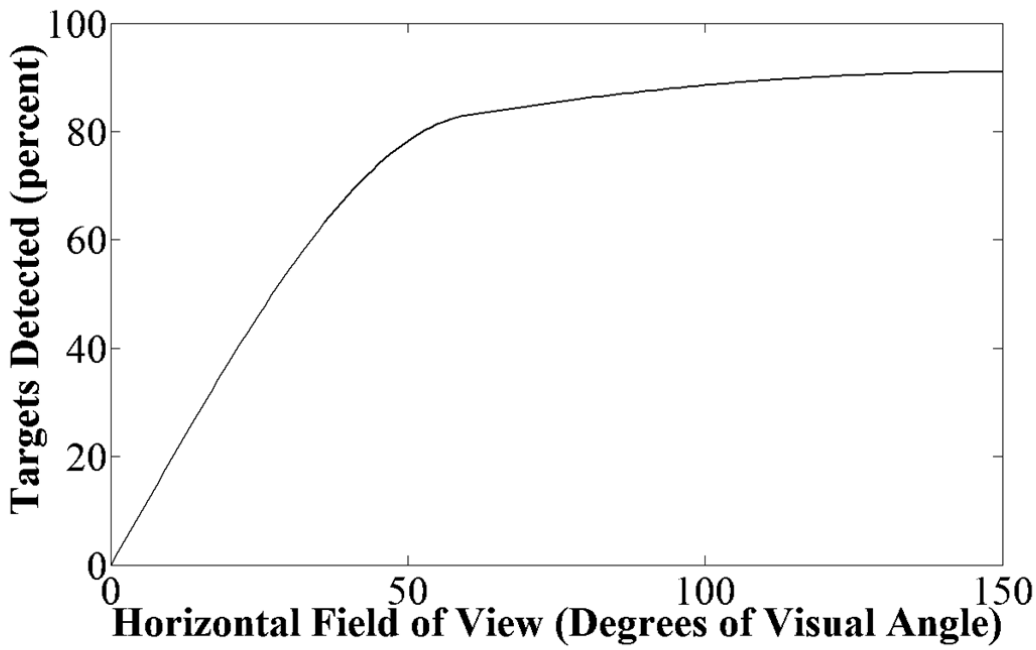


**Figure 45. Effects of FOV on Target Detection**

### *Example Human Performance Degradation: Fatigue*

In this section an example of one performance-degrading parameter ($y_n$) is provided by analyzing various studies on the impact of mass on human performance in the form of fatigue during mission lengths of time. Gallagher et al. investigated the long term fatigue effects of wearing helmets of various mass and CG on the neck of 14 male and 11 female human subjects for up to eight hours (Gallagher et al., 2008). The study measured the effects using the following

quantifiable measures as dependent variables; neck muscle fatigue via electromyography (EMG), neck strength (via Maximum Voluntary Contraction or MVC testing), neck endurance, neck discomfort surveys, and cognitive performance via a visual search task (Gallagher et al., 2008). This experiment demonstrated that overall neck strength and neck endurance measures declined significantly when comparing pre and post test measurements. Post session subject surveys indicated greater discomfort with the lighter 4.5 lb helmet with extreme forward CG shift over the heavier 6.0 lb helmet. The visual search task was meant to evaluate the effects of the extended HMD wear on cognitive performance; the hypothesis being that over the course of the eight hours of wearing the helmet the subject's performance would be degraded. To the contrary, results of the test improved over the time, likely due to learning (Gallagher et al., 2008). The authors admitted the task chosen for this study was possibly too easy and the screen size too small to generate large head movements, as there were no significant differences in the visual search results based upon helmet configuration. It should be noted that the experiment was conducted in controlled office environmental conditions which did not include the effects of actual flight such as acceleration, buffeting, vibration, and climate.

The overall human factors implications and results of this research are threefold. First, it is significant that 22 of 25 of the participants completed all five eight-hour sessions. This shows that mission lengths of this duration can be endured even in the worst case HMD design (6.0 pound helmet with forward CG shift). Second, males were observed to have more strength ($p=0.00012$) and endurance ($p=0.00845$ for the 4.5 lb, central CG configuration) than females (Gallagher et al., 2008). This helps focus investigation into human factors consideration of system development on the effects of various HMD parameters on the smaller, potentially more vulnerable populations. Finally, for all HMD applications, CG appears to matter more than mass for operator comfort. A CG-neutral helmet, if it can be achieved, seems to be better for

minimizing the head supported mass fatigue cost of the HMD system under analysis. If, however, it is necessary to place the majority of the mass forward of the natural head CG then that would have to be taken into consideration in the cost benefit analysis.

In another fatigue experiment, Eveland et al. (Eveland et al., 2008) measured neck muscle fatigue as a result of prolonged wear of weighted helmets under high acceleration levels to determine if a new, heavier variant of panoramic NVGs was more fatiguing than the legacy NVGs. In this study, subjects were under the helmet load for six hours while seated in a simulated cockpit in a centrifuge and performed mission tasks in between spurts of variable accelerative loading (never higher than 7.5 G) (Eveland et al., 2008). Results showed fatigue occurred over the course of the mission and a greater magnitude of fatigue was observed in missions with higher accelerative loading (Eveland et al., 2008). The means of the fatigue mission task performance were not statistically different, however, so it could not be concluded that the heaviest HMD configuration (6 lbs) had a greater detriment to performance than the legacy configuration (4.5 lbs) as was hypothesized (Eveland et al., 2008). Participants' survey results indicated they were most uncomfortable in the heavier helmet, but all were able to tolerate it for the entire mission simulation. The study concluded heavier helmets were tolerable and did not significantly degrade task performance in at least relatively simple cognitive tasks.

Alem et al. investigated male pilot performance while exposed to long durations of whole body vibration with variable HMD mass and CG configurations (Alem et al., 1995). The human factors metric under investigation was operator vigilance. It was observed that pilot reaction time to detect and acquire targets increased as the mass moment of the HMD increased beyond 78 N-cm (Alem et al., 1995).

Figure 46 depicts a notional relationship between performance loss due to fatigue and head supported mass. Head supported mass is an example of a performance-degrading

parameter ($y_n$) in the trade space model.  This is another example of analyzing existing human performance research to define a quantifiable performance relationship for use in the trade space analysis.  Additional research is required to determine the relationship between head supported mass and performance loss past 6 lbs.

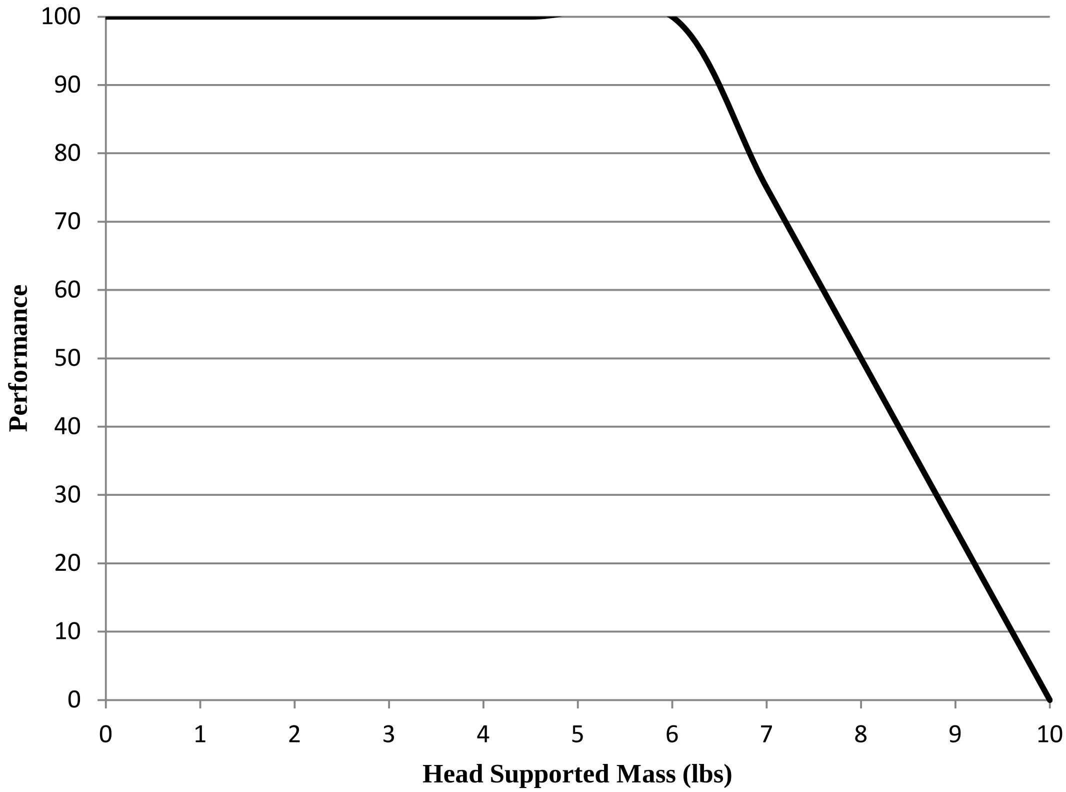


**Figure 46**.  **Generic Relationship Between Performance and Head Supported Mass**

### Example Trade Space Analysis: Quantifying HMD Contributions to TOC

Added HMD mass is likely to increase TOC as shown previously in Figure 43 (entitled "TOC in context of HMD causal loop diagram").  It is important to understand the relationship between added HMD mass and the potential increases in TOC.  In the occupational health

domain, costs come in the form of chronic neck injury to operators. In the safety domain, costs are in the form of increased acute injury. This section provides an example method of quantifying increases in TOC due to chronic and acute neck injury. The methods used to quantify acute and chronic costs could be generally applied to other aspects of HMD related TOC in a full HMD trade space analysis.

### *Occupational Health*

The AF HSI Handbook defines the occupational health domain as: "The consideration of design features that minimize risk of injury, acute and/or chronic illness, or disability, and/or reduce job performance of personnel who operate, maintain, or support the system (AF HSI Office, 2009)." For the purposes of this analysis, the occupational health domain cost of added head supported mass that will be considered is chronic neck injury. There is little documentation or literature data on the impact that increased head supported mass might have on the chronic neck injury or its long term musculoskeletal affects on users. Coakwell et al. wrote an in depth review article on the neck injury of fighter pilots (Coakwell et al., 2004). Regarding chronic neck injury, they report findings that repetitive exposure to high G forces is linked to early cervical spine degeneration (Coakwell et al., 2004). They also noted that frequent minor acute injury to the cervical spine predisposes people for more significant neck injury from trauma due to the weakening of the soft tissue supporting the spinal column (Coakwell et al., 2004). The costs of these injuries are difficult to quantify. The unknown nature of the long term effects of heavier HMDs is concerning. This could potentially be a cost to readiness if pilots are unable to fly because of chronic neck injury. It also presents an unknown long term health care cost to the government. Future study is warranted to understand these issues more fully so that this

component of the trade study can be further understood and applied within the cost benefit analysis. Equation 22 is a notional equation for the probability of chronic neck injury ($P_C$), which is formulated as a function of the exposure to high G forces (and resulting neck loading – $load_{neck}$) over time. Neck loading could be further described as a function of additional parameters if desired, to include HMD mass, HMD CG, and expected accelerative input.

$$P_C = \int_0^t f(load_{neck})dt \tag{22}$$

Equation 23 is an overall cost function for chronic neck injury incorporates the probability of chronic injury equation combined with the number of pilots in the population of interest ($n$), and the medical costs to treat the chronic injury ($C_{C\_med}$). It will be assumed for chronic injury that the pilot completes his flying career, thus there is no cost to train a replacement pilot.

$$C_C = P_C \bullet n \bullet C_{C\_med} \tag{23}$$

***Safety***

The AF HSI Handbook defines the safety domain of HSI as follows:

"The application of systems engineering and systems management in conducting hazard, safety and risk analysis in system design and development to ensure that all systems, subsystems, and their interfaces operate effectively, without sustaining failures or jeopardizing the safety and health of operators, maintainers, and the system mission (AF HSI Office, 2009)." For the purposes of this analysis, the safety domain costs of added head supported mass will include the

potential for increased injury during crash (rotary and transport aircraft), and increased injury during ejection for fighter aircraft.

Increases in head supported mass has the potential to increase the risk of acute operator neck injury if the pilot is subjected to accelerative environments, especially highly accelerative environments that can occur during ejection. Studies performed with human subjects in accelerative environments have repeatedly demonstrated significant increases in neck loads when the subjects wear an HMD than without the HMD when exposed to the same input acceleration pulse (Buhrman and Perry, 1994; Perry, 1998; Doczy et al., 2004). Injury due to a heavier HMD with an off-axis CG in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures and ligament ruptures (Buhrman and Perry, 1994). Perhaps this finding appears intuitive as increasing the mass of the head would be expected to result in an increased force when the head is exposed to acceleration. However, the human body is a complex mechanical system including a series of linkages and soft tissue connections, which have the potential to dampen an input impulse. Thus these studies have added much needed understanding on the effects of helmet mass on human neck response.

Risk curves are the foundation of an injury criterion (Pellettiere, 2012). They provide a defined relationship between neck loading and probability of injury which can be used to compare various HMD system configurations or quantify the injury risk of a prototype system during qualification or acceptance testing. A criteria tied clearly to a defined risk function allows for the acceptance of higher risk in the context of an overall system performance and cost analysis.

To quantify the safety portion of this HSI analysis, an improved pilot-scale frontal impact (Gx) AIS 2+ risk curve (Figure 47) was developed with a mathematical form similar to the National Highway Safety Transportation Administration (NHTSA) neck injury criteria

formulation called the *Nij* (Equation 24) (Eppinger et al., 2000). In Equation 4 $F_z$ is peak axial load (tension or compression), $F_{zcrit}$ is the axial load critical intercept value which normalizes the load to injury threshold based upon subject body mass, $M_Y$ is peak sagittal plane bending moment, and $M_{ycrit}$ is the bending critical intercept value. The NHTSA risk function was inadequate for application to the aviation ejection environment due to its inability to predict the 5% risk of AIS 2+ neck injury desired by the Air Force escape system oversight office, and because it has never been validated with human subject data. The improved risk curve was constructed using existing human subject testing neck data (n=67, 6G / 2kg, 8G / 1.6kg, 8G / 2kg experimental configurations) combined with cadaver data from published research (N=6, 32-39G) (Parr et al., 2013).

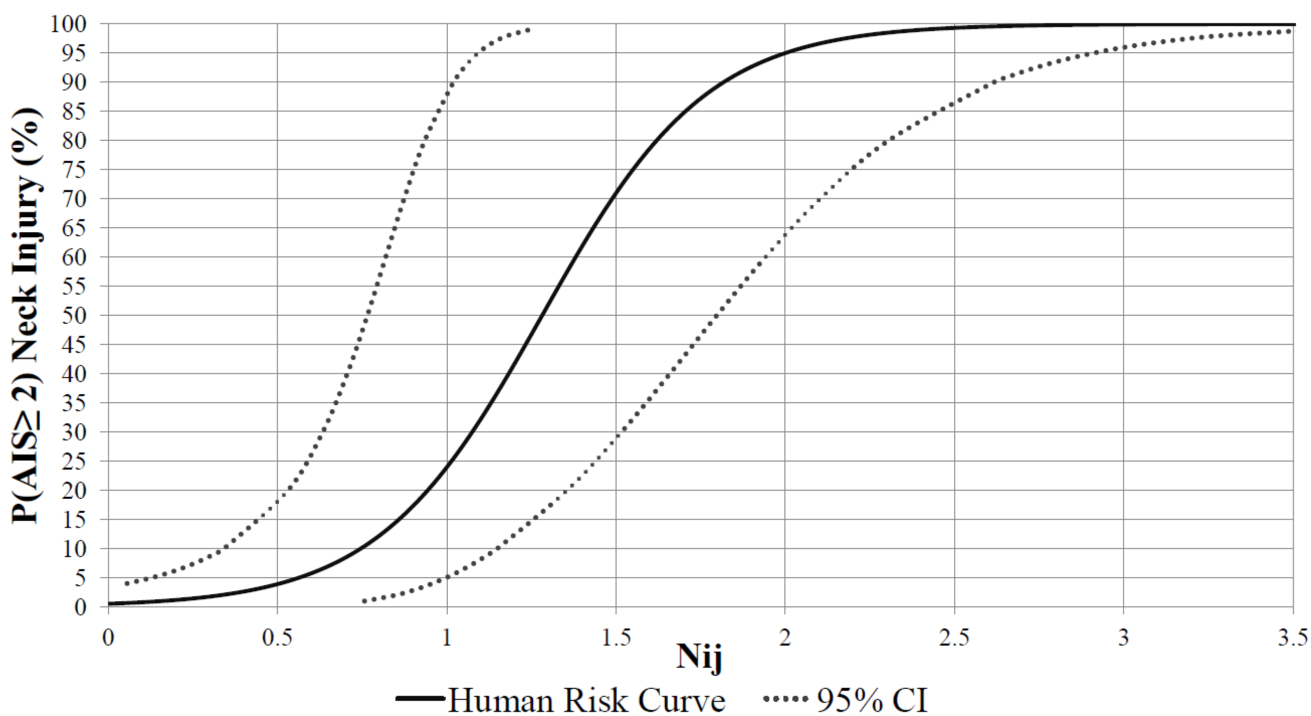


**Figure 47**. **Probability of AIS 2 or Greater Human *Nij* Neck Injury Risk Curves (95% CI Show for Human Risk Curve)** (Parr et al., 2013)

$$Nij = \left|\frac{Fz}{Fzcrit}\right| + \left|\frac{My}{Mycrit}\right| \tag{24}$$

This risk curve was applied to two additional data sets of human subject testing neck loading data, comparing the predicted injury risk of a 10G/1.4kg HMD test with a 10G/0kg HMD test. A statistically significant difference was observed in *Nij* values between the tests. The mean *Nij* for the 0kg and 1.4kg tests were 0.108 and 0.164, which predicted a 0.81% and 1.01% risk of AIS2+ neck injury respectively. The risk curve provides the ability to ascertain the difference in risk presented by different HMD mass configurations. It could also be applied to data from different accelerative loading conditions. While small, these differences in injury prediction due to HMD mass provide a basis to quantify injury risk. This approach can be applied to other data and boundary conditions from HMD systems to quantify increases in TOC based upon acute injury risk due to HMDs.

Risk curves also enable estimates to be made concerning the cost of pilot neck injury from various HMD masses on life cycle costs using historical ejection rates, probability of injury (taken from the risk curve), and pilot replacement costs. While the example above uses the *Nij* as the loading input to the risk function, any desired neck injury criteria formulation could be used to determine the probability of acute injury ($P_A$). Similar to chronic neck injury probability, the probability of acute neck injury is a function of neck load ($load_{neck}$), and neck load can be treated as a function of HMD mass, HMD CG, and acceleration. Generically we can put this probability of acute injury into the form of the Equation 25.

$$P_A = f(load_{neck}) \tag{25}$$

The probability of acute injury is incorporated into an overall cost function (Equation 26) for acute injury ($C_A$) along with the number of pilots in the applicable population of interest ($n$), the medical costs associated with the acute injury ($C_{A\_med}$) , as well as the cost to train a replacement pilot ($C_{pilot}$). The assumption will be made that an acute injury removes the pilot from any further flying duty.

$$C_A = P_A \bullet n \bullet C_{A\_med} + P_A \bullet n \bullet C_{pilot} \tag{26}$$

### Combined Performance and Cost Equations

Below are proposed equations that incorporate aggregated HSI performance and costs. Total system performance attributable to the HMD system can be quantified by the sum of each performance parameter (Equation 27).

$$p_{tot} = [p(x_1) + p(x_2) + p(x_n)] - [p(y_1) + p(y_2) + p(y_n)] \tag{27}$$

The cost equation is constructed based upon the definition of TOC. *TOC* includes the sum of HMD LCC ($LCC_{HMD}$), chronic injury costs ($C_C$), ejection injury costs ($C_A$) and other costs ($C_n$) which might be desired to include in the model (Equation 28). The major components of $LCC_{HMD}$ include research and development costs, investment costs, operating and support costs, and disposal costs (DAU, 2013a). *TOC* is minimized when each of the costs are minimized. This *TOC* equation contains only the portions of LCC applicable to the HMD system. In the portions of the equation where cost is tied to a probability ($C_C$ and $C_A$), *TOC* minimization occurs when the probabilities of chronic injury ($P_C$), acute injury ($P_A$), and other desired cost functions ($C_n$) are minimized.

$$TOC = LCC_{HMD} + C_C + C_A + C_n \tag{28}$$

Once fully described, the performance ($p_{tot}$) and *TOC* functions can be used to perform appropriate HSI optimizations. Considering the interpretation provided previously in the paper that HSI doctrine dictates maximizing performance and minimizing *TOC*, consider Equation 29 as a high level expression which aggregates the components of the trade study. The ideal HMD is one that maximizes the ratio of total performance ($p_{tot}$) to *TOC* over the parameters of $\overline{x}$ and $\overline{y}$. This expression might provide an appropriate method to consider the overall trade space of performance benefits and costs associated with the added mass of an HMD.

$$HMD_{Ideal} = max_{\overline{x},\overline{y}}\left(\frac{p_{tot}}{TOC}\right) \tag{29}$$

**Concluding Remarks**

Applying HSI doctrine to a system optimization application like the HMD design trade space is a complex undertaking which spans a multitude of research communities. Quantifying the performance benefit and lifecycle cost elements required to perform a robust HSI trade analysis will necessitate targeted research. Future work should focus on providing added understanding to each component of HMD performance and cost to more fully develop this model. In the short term, priority should be placed on creating first order approximations relating HMD system parameters to operator performance to demonstrate the extrapolation of this concept to a full model. In the longer term, targeted research should be conducted to specifically understand how common HMD system parameters such as night vision, FOV, resolution, and binocular systems influence operator performance to either validate the first order approximations or expand the functions to more robust representations.

# X. Conclusions and Recommendations

## Summary

This dissertation was motivated by the fact that acquisition programs for recently-developed, ejection-seat-equipped military aircraft have encountered delays due to the difficulty in translating pilot safety-oriented operational performance requirements into robust verification criteria. Specifically, the operational requirement that a pilot not experience greater than a 5% risk of an AIS 2 level injury during ejection has not previously been translated to clear and defendable system level requirements. As a result, qualification tests have produced results where the implications of these results have been highly debated by the stakeholders across the acquisition and contractor communities. These debates have absorbed resources and potentially resulted in late stage design changes that can be expensive and time consuming to implement. This issue has become especially important as the ejection systems, the head-mounted mass in the form of HMDs, and the anthropometrics of the pilot population are undergoing simultaneous change, all of which likely affect the safety of the pilot during ejection.

To address this problem, the current dissertation attempted to define this problem and review the motivation for the current work, including unique risks posed by head supported mass in accelerative environments as described in Chapter I. Chapter II provided an overview of existing neck injury criteria and explored the fundamentals necessary to construct improved, aviation specific, neck injury criteria for use as the foundation for future Air Force qualification testing criteria. A method for developing new injury risk curves in each of the three major axes of acceleration using the most appropriate neck load input function and statistical methods combined with existing human and PMHS data sets was demonstrated in Chapter III. This method was applied to derive an initial set of criteria and new injury risk curves, one for each of

the three major axes of acceleration (Gx, Gy, and Gz) in Chapter IV, V, and VI. These new criteria and their respective neck injury curves then form the basis of a set of new pilot-scale, multi-axial neck injury criteria, applicable to assessing ejection system safety. These three axis-specific (Gx, Gy, Gz) sub-criteria make up the complete MANIC and are summarized in Chapter VII. The new pilot-scale MANIC was then applied to two existing data sets of ejection system tests as a feasibility case study to compare the MANIC with legacy criteria to make a preliminary assessment of the new criteria. Chapter VII also identified future data needs to enable the development of a more comprehensive MANIC. Chapter VIII incorporated the actors and data involved across the neck injury criteria enterprise into the DoD Architecture Framework, helping to provide insight into the system and illuminate potential system interface issues involved in developing, implementing, testing, and complying with the requirements of USAF escape system neck injury criteria. Finally, Chapter IX described the development of a method for the formulation of a quantitative, first order, trade space model of parameters influencing the HSI impacts of HMDs. This model explored how a set of robust safety criteria, such as the MANIC, could be incorporated with information from other HSI domains, including human factors engineering, survivability, fatigue, and occupational health to demonstrate how quantifying applicable elements of these domains might result in progress toward accomplishing the HSI goal of maximizing system capability and minimizing TOC for an HMD system.

The methods proposed to develop improved neck injury criteria detailed in this work will help evaluate risk posed by various HMD and escape system configurations; they will also provide decision makers with a quantitative method to conduct trade studies in the safety domain during the HMD acquisition process. It is also anticipated that the pilot-scale MANIC may be used as a foundation for an overall aircraft escape system developmental testing standard used in the acquisition process to ensure pilot neck safety. Additionally, as the research objectives were

pursued, gaps in the existing body of human and PMHS data were discovered. Data deficiencies were identified; these will need to be addressed by future research and incorporated into the MANIC developed here to achieve fully supported, statistically robust multi-axial aviation specific neck injury criteria. It is hoped that the development of this new data driven family of criteria with rigorously developed neck injury curves will aid the translation of pilot safety-oriented operational performance requirements into robust system verification criteria to aid the development of future ejection safety systems and the associated components.

**Conclusions of Research**

There are a number of important conclusions that arose as a result of this research topic. First, the combined human and PMHS data method for risk function construction has both advantages and disadvantages in the current environment. The primary advantage is that it captures aggregate human tolerance to accelerative loading, accounting for both ends of the injury spectrum (human subject tolerances and injury thresholds). Unfortunately, this method requires the estimation of neck loads from calculations using head acceleration. While it is believed that this method yields usable information, it is not as exact as instrumented ATD observed loading. Further, this method requires the assumption that the Hybrid III ATD neck is biofidelic in order to directly apply criteria to ATD for system evaluation or the derivation of transfer functions, which was outside the scope of the current dissertation. This method also has the disadvantage that the injury data, obtained from PMHS, is from passive subjects, which come from a diversity of ages. Therefore, the PMHS are potentially unrepresentative of the healthy, fit, younger pilot population. Other issues include the limited amount of data that exists due to the small sample size PMHS injurious data sets. These small sample sizes, the assumed structure

of criteria, and the number of categories of critical values all impact risk function development, potentially leading to less than optimal solutions.

Based upon the existing literature and observations of comparisons of neck injury risk functions constructed with the range of statistical tools available, survival analysis was found to be an important statistical technique to use with censored human/PMHS data, as compared to the legacy method of logistic regression. Since both the human subject and PMHS data is not exact, this data routinely violates the assumptions of logistic regression. Survival analysis appears to provide relatively robust solutions when applied to small data sets, as long as overlap exists between the non-injury and injury data sets.

**Significance of Research**

Existing neck injury criteria are either not adequately tailored for the requirements of military aviation and the ejection environment or are too cumbersome to be used to evaluate and perform trade studies between added capability and safety on escape systems incorporating HMDs. Existing neck injury criteria also allow for higher injury risk and higher levels of acceptable injury than currently desired by the USAF escape system community. The MANIC developed in this research is a state of the art human injury criterion. New in the approach detailed in this research is a method of developing risk functions, based upon human subject and PMHS data, tailored to the specified requirements of the USAF escape community. Assuming proper ATD transfer functions can be created, this research provides a scientifically valid path to a set of test criteria capable of linking stakeholder safety requirements to measureable system level requirements, and may serve as the basis for new USAF developmental testing criteria. The development of improved USAF aviation-specific, ejection neck injury criteria might

provide the necessary basis for the acquisition community, to more accurately and adequately assess the safety of escape systems incorporating HMDs. Also new in this approach is the attempt to quantify the safety impacts of added head supported mass and apply this to a quantitative HSI model of the trades between HMD safety and capability. Additionally, improved criteria could provide decision makers from the design and user communities with information on the risks of neck injury along the spectrum of neck loads, providing a method to more fully balance risk exposure with potential operational benefits that come from added capabilities. This method, as well as a set of specific criteria for the HMD, is valuable to the acquisition and escape community.

**Recommendations for Future Research**

There are a number of ways future research can add to the methods and the pilot scale MANIC developed in this research. Injury data from PMHS testing is most important to risk function development and also the most lacking. Specifically, additional -Gx PMHS experimental neck data (6-load and AIS injury) are needed. Also, +Gz (vertical impact) PMHS combined loading neck data (6-load and AIS injury) is needed to capture neck injury mechanisms and limit loading in the catapult phase of ejection. Another significant data issue that should be addressed in future experiments was the fact that +Gz human subject tests in the AFRL biodynamic data base did not observe x- or z-axis head rotational acceleration data needed to determine *Mx* and *Mz* for the upper neck, and Gy tests did not observe x-axis head rotational acceleration data to determine *Mx*. This would result in the elimination of *Mx* or *Mz* from use in future +Gz injury criteria development and resulted in the elimination of *Mx* from use in the Gy injury criterion development described in Chapter VI. Future experiments should capture all six

major upper neck loads (*Fx*, *Fy*, *Fz*, *Mx*, *My*, and *Mz*) if possible. Incorporating human subject and PMHS data from specifically designed future experiments into the criteria developed in this research will add necessary robustness to the risk functions and resulting criteria. Additionally, research into a more cost effective and feasible means of obtaining real world injury data is needed, such as collection data from existing accidents through the incorporation of high speed cameras in select automobiles or through the use of micro-accelerometers on pilot helmets. Research into developing human to ATD transfer functions to translate the human based criteria developed in this study to ATD data is the next step toward a robust, multi-axial, AF escape system testing criteria.

Methods for determining optimal critical values are needed. Results from this dissertation research demonstrated that the critical values did not adequately normalize for body mass when used in a combined loading formulation in y-axis accelerative experiments and thus are lacking and need improvement. Furthermore, additional research is needed to identify the error due to assumptions in the present criteria, such as the application of a model-validating data set.

**Summary**

The introduction of HMDs into the aircraft escape system poses new and significant Human Systems Integration design, development, and test concerns. These concerns are especially severe for individuals with smaller anthropometric dimensions who may be more susceptible to injury due to the larger added helmet mass to head mass ratio. Where previously no adequate USAF neck injury criteria existed to effectively guide the development and test of current and future HMD-centric escape systems, this research has contributed to filling that void

by presenting a novel method to develop neck injury criteria. The updated pilot-scale injury criteria risk functions are constructed with combined human subject and post mortem human subject experimental data using a parametric survival analysis. The MANIC demonstrated sensitivity to real world data and demonstrated the ability to limit specific injury risk levels at user determined AIS injury classification levels, capability not available in legacy criteria. The MANIC risk functions may provide the foundation for future, formalized USAF neck injury criteria which, when applied to developmental and qualification testing of escape systems, will ensure pilot safety and limit risk of neck injury.

## Appendix A: SAFE Association Conference Paper – A Human Systems Integration Analysis of Helmet Mounted Displays


### Abstract

Helmet mounted displays (HMDs) provide increased capability to advanced aircraft systems but also add mass to the pilot's head. This mass potentially increases fatigue, degrades pilot scan patterns, and potentially increases chronic, as well as acute injury during accelerative loading. From a Human Systems Integration (HSI) perspective, HMD capabilities should be selected to maximize performance and minimize system total ownership costs (TOC). Unfortunately, a clear method does not exist for performing this HSI tradeoff analysis to include safety (acute neck injury), occupational health (chronic neck injury), human factors engineering (performance and fatigue), and survivability. This study utilized content analysis and data to develop a qualitative model of the impacts HMDs have on HSI. Further, recent research on neck injury risk criteria was applied to quantify the impacts of helmet mass on the ejection safety portion of the model. A methodology for the formulation of a quantitative model of parameters influencing the HSI impacts of HMDs was developed. This study illustrates the difficulty in formulating a rigorous optimization of HSI parameters for a HMD. If quantitative HSI assessments of realistic system performance and TOC are to be conducted, additional research will be required.


### Background

Human Systems Integration (HSI) is a method for addressing human-centered concerns during system design. Department of Defense Instruction (DoDI) 5000.02 directs that HSI is to be applied to "optimize total system performance, minimize total ownership cost" and insure that the system accommodates the user population (DoD, 2008). However, "optimizing total system performance" may not be consistent with "minimizing Total Ownership Costs" (TOC). As a result, this criterion might be reinterpreted to maximizing the ratio of total system performance to TOC. Other possible optimization formulations consist of either maximizing total system performance subject to some maximum TOC constraint, or minimizing TOC subject to a total system performance minimum constraint. Regardless of this interpretation, it is first necessary to quantify each of these attributes to enable "optimization" (e.g., minimization or maximization).

HSI has gained emphasis, both within military acquisition (Booher, 2003) and the systems engineering community (Madni, 2009). The HSI concept assumes that by associating human-centered concerns with human-centered domains one can arrive at an improved system solution. This solution considers the impact of these concerns within each domain and synthesizes the results to understand the impact of potential system trades on total system performance and TOC. These domains, which often include manpower, personnel, training,

human factors, occupational health, and safety, represent areas of human-centered technical expertise which ideally can assess the impact of system trades on performance and cost (AF HSI Office, 2009). This paper addresses the question of whether optimization can be achieved as instructed by DoDI 5000.02 for human-centered systems and proposes a methodology for the formulation of a quantitative model of parameters to perform HSI trade space analysis on HMD systems.

To address this question, an analysis is applied to the design of helmet mounted display (HMD) systems for fixed-wing aircraft. Within the human factors literature, it has been documented that the use of HMDs can improve pilot situation awareness (SA) as critical information can be displayed to the user without requiring visual search or fixation on head-down displays (Geiselman and Havig, 2011). However, as with many technologies the intended technical improvements often have negative unintended consequences. For example, such a display may increase the head supported mass to values beyond 5 lbs. Such an increase in head supported mass for the entire mission duration is a significant departure from the operational procedures of legacy systems, and the totality of costs and performance are not well understood.

HMDs were selected for this analysis as they are used in numerous legacy fixed and rotary wing aircraft and are likely to be common human-machine interface equipment in the future of manned flight. Currently HMDs are in use on multiple Department of Defense (DoD) weapon systems (e.g., AH-64, F-15, F-16, F-18) and planned for the F-35 Joint Strike Fighter. In fact, this human interface technology is becoming more prevalent with increasing capability (VSI, 2013). Throughout the remainder of this paper it will be assumed that this trade space model for evaluation of HMD systems could be applied to a range of military applications including fixed wing, rotary wing, and mounted or dismounted ground operations.

The benefits of addressing HSI domains early in the systems acquisition lifecycle have been documented (Booher, 2003; INCOSE, 2011); however, these documented examples rely heavily on expert opinion rather than rigorous quantitative trade analysis. Limited quantitative HSI research exists in the literature, though Hardman has put forth robust HSI engineering methodologies in the areas of aircraft mishap prevention requirements and user interface design (Hardman, 2009). Numerous human factors studies have been performed which attempt to explain and quantify human visual and cognitive performance relative to military applications of HMDs (Rash et al., 2009), but this research has not been adequately translated into an overarching quantitative HSI application. This example illustrates the difficulty which arise when attempting to provide a quantitative HSI analysis within a practical (although constrained) systems engineering process.

**Applicable Definitions**

As HSI is a multidisciplinary field, it is important to specifically define important terminology. First, monetary cost terms to clearly differentiate include Lifecycle Cost (LCC) and Total Ownership Cost (TOC). Both the DoD and the International Council on Systems

Engineering (INCOSE) define LCC as the totality of acquisition and ownership costs of a system over its entire life to include concept, development, production, operation, sustainment and disposal (DAU, 2013a; INCOSE, 2011). LCC also includes indirect costs that can be reasonably linked to the system. TOC incorporates LCC, but also includes “related infrastructure or business processes costs not necessarily attributed to the program in the context of the defense acquisition system” to include medical care, which is especially germane to the current study. INCOSE provides less distinction, but perhaps allows for more flexibility, between its definition of LCC and TOC, incorporating into TOC some of the costs which the DoD considers in its definition of LCC (INCOSE, 2011). In general, however, INCOSE considers much of the HSI related costs in its definition of TOC, including personnel costs, training costs, costs of mishaps, and disability compensation and liability claims (INCOSE, 2011). What is important for the purposes of this paper is to establish that we aim to minimize TOC, which includes LCC plus medical care costs associated with the HMD system that may extend beyond the lifecycle of the program.

Although the research and development components of LCC are closely monitored during the acquisition lifecycle and unexpected expenses incurred during this phase of a system’s lifecycle draws significant public scrutiny and compensatory legislation (WSARA, 2009), the operation and support components of LCC have more recently began to draw similar scrutiny, although projection of these costs are more difficult (Ryan, 2012). Total ownership costs have received less focus than either acquisition or lifecycle costs as these costs can arise from unpredictable sources such as environmental contamination by an unknown carcinogen or other human systems hazards with consequences that are unknown or difficult to project.

While monetary costs are an important consideration within the current paper, it is also understood that system attributes intended to improve the performance of the operator or the system can additionally reduce the performance of one or more of these entities and therefore cost can also refer to loss of performance. Total System Performance refers to the quantifiable mission capability performance of the system to which the HMD contributes. In this study the aspects of total system performance not related to the HMD will be assumed to be constant and will not be considered since it is desired to analyze only the performance contributions of the HMD system. Operator Performance (in Figure 1) is specifically the cognitive, sensory, and physical human performance that is either enhanced or degraded by the HMD.

## Outlining the Trade Space

A causal loop diagram was created, as shown in Figure 1, to depict a portion of the relationships affecting HMD utility from an HSI point of view. The traditional Systems Engineering (SE) top down functional decomposition begins with capabilities (operational requirements) from which system requirements are generated. System functions are generated from system requirements and then allocated to individual components which are described by parameters. In Figure 1, the “HMD System Parameters” block represents those parameters which were allocated to the HMD.

As shown in Figure 1, as HMD system parameters increase, the performance of the operator should be expected to increase. In the diagram, a plus sign signifies that a change in the first entity causes a change in the same direction in the second entity, while a minus sign signifies that a change in the first entity results in a change in the opposite direction of the second entity. Improvements in operator performance would be expected to improve overall system performance. Improvements in operator performance will also likely reduce the probability that the pilot will need to eject from the aircraft, improving system survivability, which, in turn, may reduce the likelihood of acute neck injury during ejection from the aircraft.

This proposed increase in HMD system parameters often requires modification of HMD system hardware, which can increase the mass of the HMD. This mass is supported by and adds load to the human operator's neck and spine. This increase in mass then tends to increase the fatigue of the operator, which is often considered within the Human Factors Engineering domain. Further, the likelihood of chronic neck injury occurring due to the repeated exposure of the neck to greater than natural forces as the pilot is exposed to accelerative environments, such as those posed by vibration, buffeting, or high rate maneuvers also increases. Chronic neck injury is often considered within the domain of occupational health. Finally, as the mass of the HMD increases, the likelihood and severity of acute neck injury also increases as the pilot may be exposed to high accelerations, for example during ejection. This effect of mass impacts the Safety domain. Note that each of these factors has the potential to decrease operator performance. Fatigue directly impacts the performance of the operator throughout the mission while chronic neck injuries and safety issues will likely reduce the time that an operator can perform within the platform increasing the costs for operator recruiting and training while reducing average operator experience. Therefore, the platforms will, on average, have less experienced operators who may have a lower performance than more experienced operators. Increases in operator performance are expected to increase total system performance. While it is understood that total system performance involves more than just the contribution of the operator's performance which is enhanced through HMD system parameters, the current analysis is limited only to this portion of total system performance.

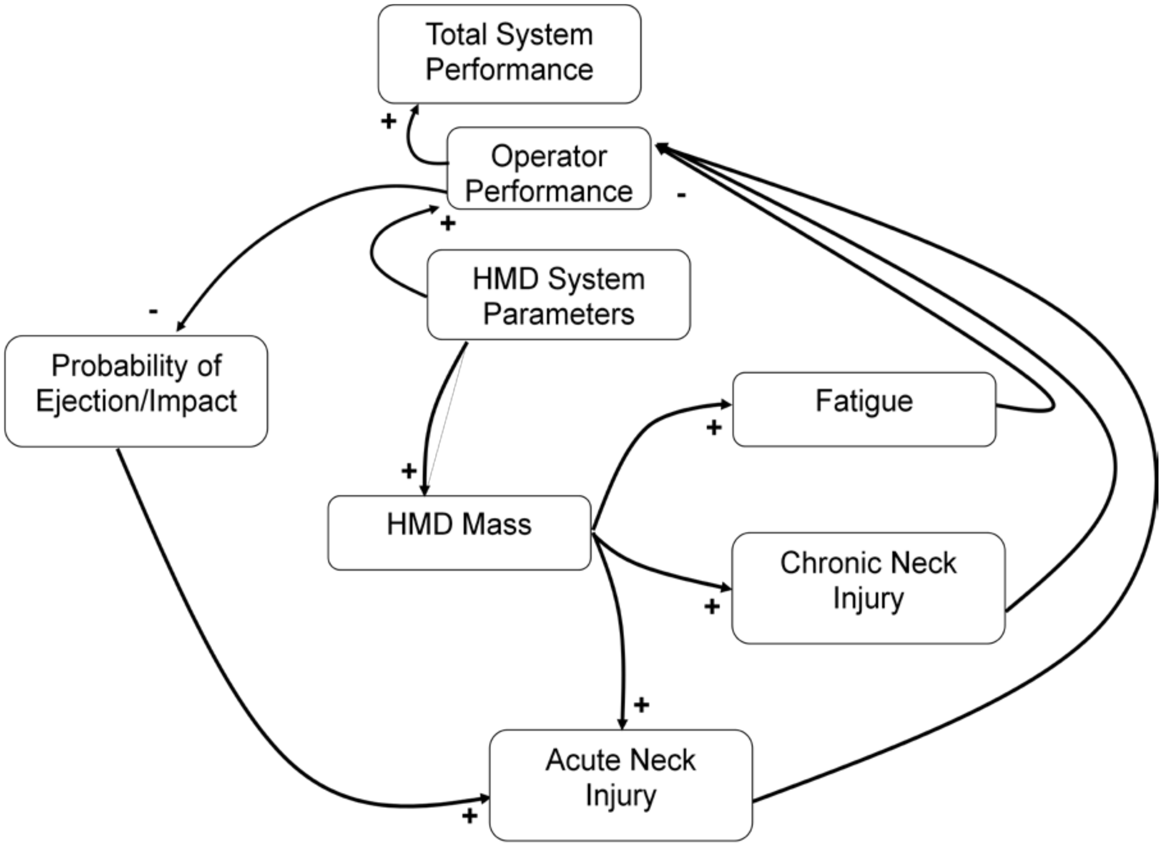


**Figure 1: Partial HMD HSI causal loop diagram**

During technology development, effort may be spent to reduce the effect of increases in HMD system parameters on HMD Mass through the use of lighter materials, increases in technology integration, or other technological innovations. However, technology development requires investment, increasing development costs, which may increase or decrease TOC, as depicted in Figure 2. It is possible that the investment in HMD developmental costs during acquisition may decrease HMD mass which would decrease the fatigue and chronic/acute neck injury costs associated with the additional mass. Assuming added HMD system parameters increase HMD mass, TOC is likely to increase as acute and chronic neck injuries increase. Decreases in the probability of ejection will likely decrease TOC. Therefore, it is intuitive that tradeoffs exist within the HMD design which influence total system performance and TOC.

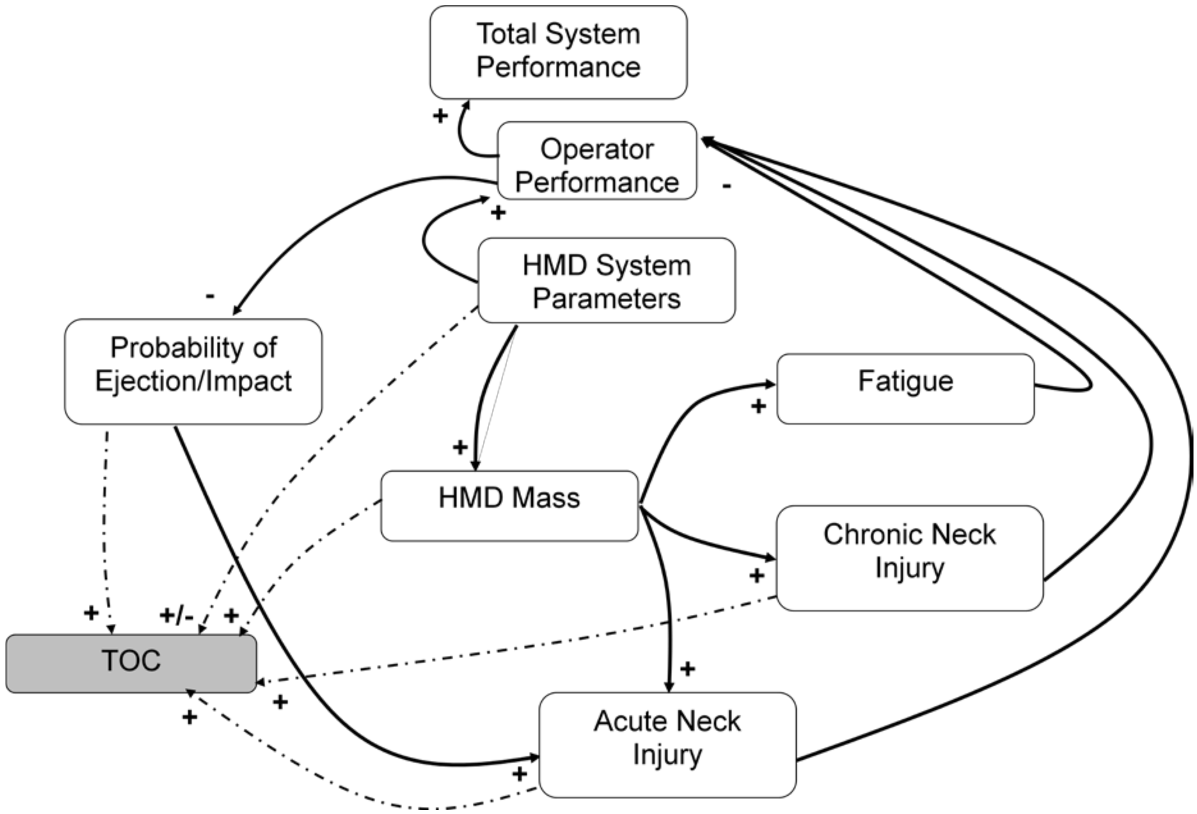


**Figure 2: TOC in context of the partial HMD causal loop diagram**

Although Figure 2reflects a number of high level trades with respect to HMDs, it does not reflect any specific changes in HMD system parameters or the impact of these parameters on performance. As depicted in Figure 44, this added capability could, for example, include changes to HMD parameters. For instance design parameters of the HMD could include field of view (FOV), night vision, resolution, and binocular or monocular viewing. Each of these parameters may affect human performance and, therefore, total system performance. However, these relationships are often not well specified. Instead, the human factors literature will often associate changes in these HMD parameters with intermediate attributes, such as the ability of an operator to search for and detect a target, target the enemy, detect or understand platform motion, maintain comfort without eye strain, or determine the orientation of their aircraft, as shown in Figure 3. While performance on these tasks is likely to influence operator performance and therefore total system performance, this relationship is often difficult to ascertain.

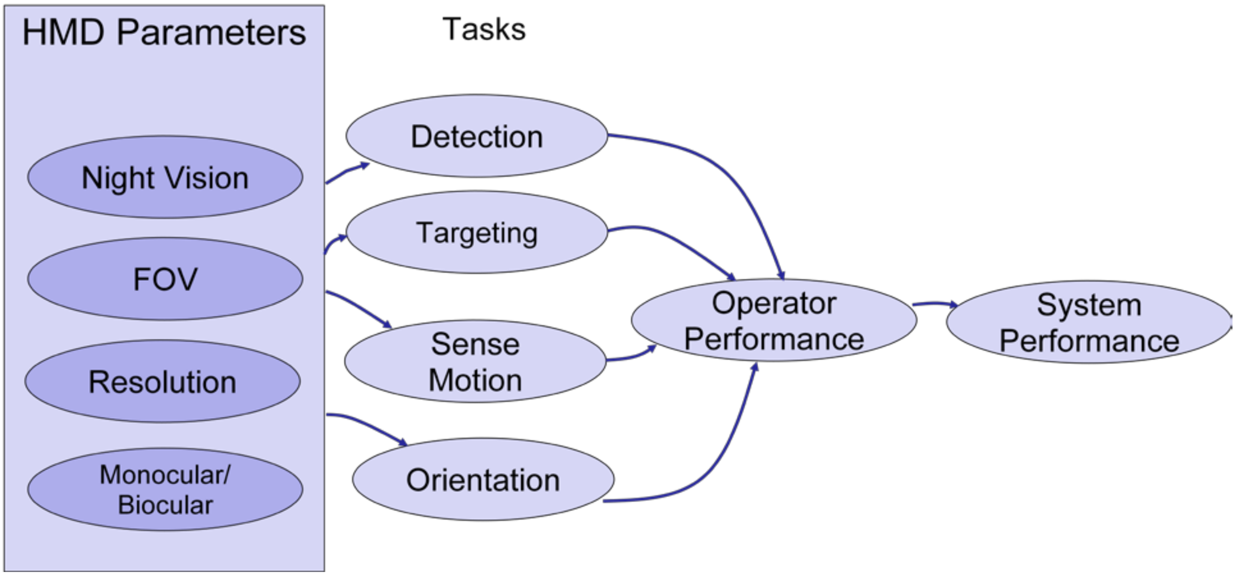


**Figure 3: HMD parameters and tasks that contribute to system performance**

In an analysis of the system, it is assumed that increases in HMD system parameters, such as the night vision and biocular optics are being added to increase operator performance and as a result, improve total system performance. Within the context and scope of HSI, however, increased operator performance comes at a cost, not only in monetary terms, but potentially in total system performance as indicated in Figure 2. For example, the cost is an increase in HMD mass, which is likely to increase fatigue over mission length durations, decreasing operator performance. These unintended costs and negative impacts on operator performance must also be fully understood to quantify HSI tradeoffs. Both the benefits and the costs to human performance, safety, and health must be identified at the earliest point possible in system development to make the appropriate HSI cost and benefit trade decisions in the system design. Table 1 provides a summary of the HSI domains with costs and benefits in the HMD system trade space.

**Table 1: HSI Domain Cost/Benefit Comparison**

| HSI Domain | Human Performance Benefit | Human Performance Cost |
|---|---|---|
| Human Factors | SA, Target Cueing, Precision Navigation | Performance degradation due to neck fatigue |
| Survivability | Increased due to improved SA, Target Cueing, Precision Navigation | |
| Safety | | Increased risk of acute neck injury during ejection/crash |
| Occupational Health | | Increased risk of chronic neck injury |

## Human Performance:  Overview of Applicable HSI Domains

The benefit from adding HMD capability lies primarily in the HSI domains of human factors engineering and survivability.  In the human factors engineering domain, benefits include increased SA, increased target cueing capability, and increased precision navigation capability. In the survivability domain, increased performance is realized from the previously mentioned human factors benefits (SA, target cueing, and precision navigation) each contributing to human performance, which increases survivability of the weapon system, thus increasing total system performance.

Incorporating HMD system parameters usually requires adding mass to the unit, which increases the operator's head supported mass.  The cost to operator performance of this additional head supported mass comes in the HSI domains of survivability and human factors.  In the human factors domain, performance degradation comes in the form of neck fatigue caused by HMD mass potentially impacting the mental performance of the operator.

### Human Factors Engineering

The AF HSI Handbook defines the human factors engineering domain of HSI as follows:

"The comprehensive integration of human capabilities and limitations (cognitive, physical, sensory, and team dynamic) into systems design, to optimize human interfaces to facilitate human performance in training operation, maintenance, support and sustainment of a system (AF HSI Office, 2009)."  This section analyzes the applicable beneficial components of human factors engineering applicable to HMD system parameters.

*Situation Awareness.*  According to Rash et al., the chief objective of HMD designers is to maximize SA for the operator (Rash et al., 2009).  Endsley has put forth a widely accepted three level definition of SA as "Level 1) the perception of the elements in the environment within a volume of time and space, Level 2) the comprehension of their meaning, and Level 3) the projection of their status in the near future (Endsley, 1995)."  An Air Force definition of SA has been proposed by Geiselman as "A pilot's continuous perception of self and aircraft in relation to the dynamics of flight, threats, and mission, and the capability to forecast, then execute tasks based on the perception (Geiselman and Osgood, 1994)."  Many technologies can be added to a HMD which would provide increased SA.  Within a HMD, space for important symbology for system operation and user SA is at a premium.  Increased resolution and FOV would help alleviate this problem.  Geiselman has suggested that if additional information (specifically ownship status symbology) could be presented it could add "operational utility of the HMD by increasing lethality and survivability for day, night, and all weather application (Geiselman, 1999)."  While more information is not always better, it will be assumed in this analysis that the

presentation will be designed in accordance with established human factors practices so as not to confuse or overwhelm the pilot's ability to obtain the necessary information from the display.

*Target Cueing.* The improved performance of HMDs gives tactical fighter aircraft a distinct advantage in targeting. According to Rash et al., "HMDs are 'must have' equipment on GEN-4 fighter aircraft, since high off-boresight weapons and visual cueing outweigh any aircraft-performance advantage during a dogfight (Rash et al., 2009)." A pilot's ability to look and target with the HMD instead of with the nose of the aircraft, subjecting him/herself and the airplane to high G loading dramatically altered fighter pilot tactics, significantly increasing operator performance and total system performance. This same technology is incorporated into rotary wing HMDs for target cuing.

*Navigation.* An improved HMD with increased FOV, resolution, night vision and binocular capability would increase precision navigation performance. The flight information required for navigation could be better displayed and would allow the operator to better fuse navigation inputs thus improving this portion of the mental workload required during flight. HMDs allow the user to monitor important data without switching their visual attention from the operational environment to view the instrument panel, and then integrating information from the two disparate sources. Overall operator performance improves when key flight information is presented within the pilot's line of sight (Rash et al., 2009). Pilots are able to detect changes within their field of view since the HMD allows them to keep their gaze forward (Rash et al., 2009). A well designed layout of the navigation information within the display area will enhance human performance in this area. Additionally, night vision would enable this same capability to be leveraged at night.

**Survivability**

The AF HSI handbook defines the survivability domain of HSI as "The ability of a system, including its operators, maintainers and sustainers to withstand the risk of damage, injury, loss of mission capability or destruction. Survivability includes the elements of susceptibility, vulnerability, recoverability, and suitability (AF HSI Office, 2009)."

Many of the capability enhancements discussed previously in the human factors section, including SA, target cueing, and precision navigation also contribute to increased survivability as increases in operator performance will likely reduce the probability of platform loss. Increased SA is likely to reduce human error which could result in controlled flight into terrain, runway incursions, or mid-air collisions. For ground operators, increased SA prevents fratricide and provides increased overall battle space awareness, potentially preventing the enemy from becoming a destructive threat. Improved target cueing counters the adversary, improving blue force survivability. Precision navigation enhances maneuverability in low level terrain, specifically in rotary wing and tactical airlift operations, decreasing platform visibility.

**Example Trade Space Analysis: Human Performance**

Development of a preliminary model begins with identifying quantifiable performance trade space. In this section, the influence of an example HMD system function on operator performance is explored, and a notional or approximate relationship is shown. For a fully developed model, the user could follow this methodology for the specific HMD system parameters of interest for their specific HMD trade space analysis.

The HMD components which add mass as well as influence operator performance (*p*) are described in the formulation below by the aggregation of performance-increasing parameters ($\overline{x}$) (e.g. increased field of view, resolution, night vision, biocular system versus a monocular system) and performance-degrading parameters ($\overline{y}$) (e.g. HMD mass, HMD center of gravity). This aggregation can be stated as:

$$p_{tot} = f[p(\overline{x})] - g[p(\overline{y})]$$

$$\text{where } \overline{x} = \{x_1, x_2, ..., x_n\} \text{ and } \overline{y} = \{y_1, y_2, ..., y_n\}$$

Example Human Performance Benefit: Field of View

As an example of how one performance-increasing parameter ($x_n$) is quantified, a study linking HMD display FOV was analyzed. An element of SA is target detection. Nelson and colleagues explored the effects of FOV on operator performance, specifically detection of an oncoming aircraft. The results are shown in Table 2 (Nelson et al., 1998). Operator target detection performance was observed to increase as a function of FOV.

**Table 2: Target detection as a function of field of view**

| Field of View | Correct Detection (%) | Detection Distance (m) |
|---|---|---|
| 60x40 | 83 | 1800 |
| 150x70 | 91 | 2150 |

This data can be used to derive the model depicted in Figure 4. Although this model approximates the impact of FOV on human target detection, the relationship between this function and total system performance would require further study. Using a similar methodology, research from other human performance studies could similarly define the link shown in Figure 3 between HMD parameters, operator tasks, and operator performance. Studies like the Nelson el al. research provide a quantifiable link for use in an overall HMD trade space analysis.

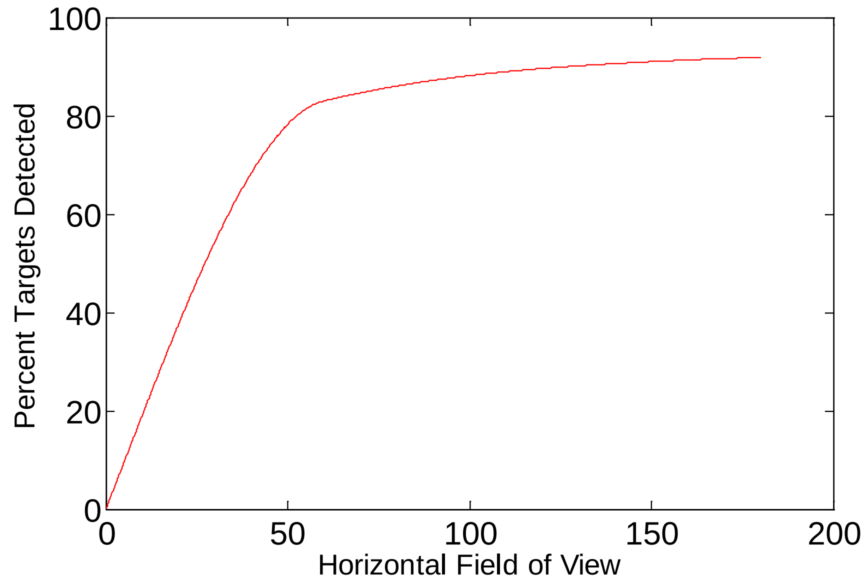


**Figure 4: Effects of FOV on Target Detection**

Example Human Performance Degradation: Fatigue

In this section an example of one performance-degrading parameter ($y_n$) is provided by analyzing various studies on the impact of mass on human performance in the form of fatigue during mission lengths of time. Gallagher et al. investigated the long term fatigue effects of wearing helmets of various mass and CG on the neck of human subjects for up to eight hours (Gallagher et al., 2008). The study measured the effects using the following quantifiable measures as dependent variables; neck muscle fatigue via electromyography (EMG), neck strength (via Maximum Voluntary Contraction or MVC testing), neck endurance, neck discomfort surveys, and cognitive performance via a visual search task (Gallagher et al., 2008). This experiment demonstrated that overall neck strength and neck endurance measures declined significantly when comparing pre and post test measurements. Post session subject surveys indicated greater discomfort with the lighter 4.5 lb helmet with extreme forward CD shift over the heavier 6.0 lb helmet. The visual search task was meant to evaluate the effects of the extended HMD wear on cognitive performance; the hypothesis being that over the course of the eight hours wearing the helmet the subject's performance would be degraded. To the contrary, results of the test improved over the time, likely due to learning (Gallagher et al., 2008). The authors admitted the task chosen for this study was possibly too easy and the screen size too small to generate large head movements, as there were no significant differences in the visual search results based upon helmet configuration. The overall human factors implications and results of this research are threefold. First, it is significant that 22 of 25 of the participants completed all five eight-hour sessions. This shows that mission lengths of this duration can be

endured even in the worst case HMD design (6.0 pound helmet with extreme forward CG shift). Second, males were observed to have more strength and endurance than females. This helps focus investigation into human factors consideration of system development on the effects of various HMD parameters on the smaller, potentially more vulnerable populations. Finally, for all HMD applications, CG appears to matter more than mass for operator comfort. A CG-neutral helmet, if it can be achieved, seems to be better for minimizing the head supported mass fatigue cost of the HMD system under analysis. If, however, it is necessary to place the majority of the mass forward of the natural head center of gravity then that would have to be taken into consideration in the cost benefit analysis.

In another fatigue experiment, Eveland et al. measured neck muscle fatigue as a result of prolonged wear of weighted helmets under high acceleration levels to determine if a new, heavier variant of panoramic NVGs was more fatiguing than the legacy NVGs (Eveland et al., 2008). In this study, subjects were under the helmet load for six hours while seated in a simulated cockpit in a centrifuge and performed mission tasks in between spurts of variable accelerative loading (never higher than 7.5 G) (Eveland et al., 2008). Results showed fatigue occurred over the course of the mission and a greater magnitude of fatigue was observed in missions with higher accelerative loading (Eveland et al., 2008). The means of the fatigue mission task performance were not statistically different, however, so it could not be concluded that the heaviest HMD configuration (6 lbs) had a greater detriment to performance than the legacy configuration (4.5 lbs) as was hypothesized (Eveland et al., 2008). Participants' survey results indicated they were most uncomfortable in the heavier helmet, but all were able to tolerate it for the entire mission simulation. The study concluded heavier helmets were tolerable and did not significantly degrade task performance in at least relatively simple cognitive tasks.

Alem et al. investigated male pilot performance while exposed to long durations of whole body vibration with variable HMD mass and center of gravity (CG) configurations (Alem et al., 1995). The human factors metric under investigation was operator vigilance. It was observed that pilot reaction time to detect and acquire targets increased as the mass moment of the HMD increased beyond 78 N-cm (Alem et al., 1995).

Figure 5 depicts a notional relationship between performance loss due to fatigue and head supported mass. Head supported mass is an example of a performance-degrading parameter ($y_n$) in the trade space model. This is another example of analyzing existing human performance research to define a quantifiable performance relationship for use in the trade space analysis. Additional research is required to determine the relationship between head supported mass and performance loss past 6 lbs.

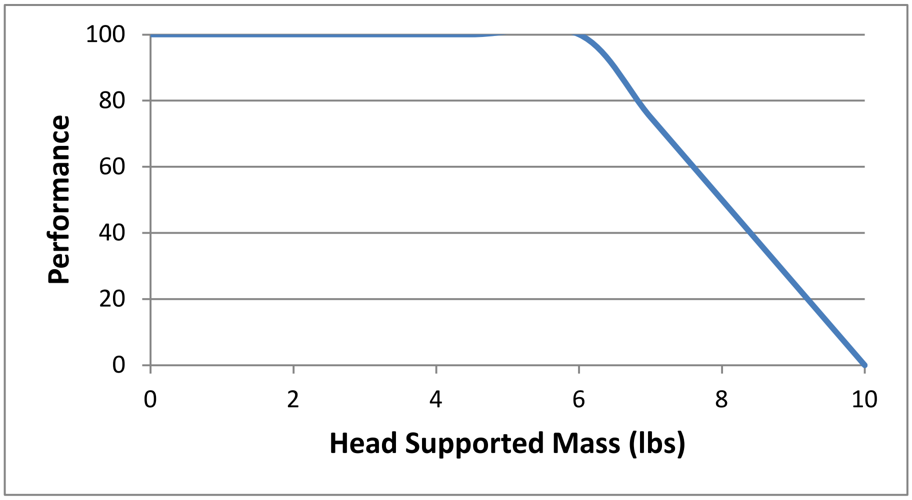


**Figure 5: Generic relationship between performance and head supported mass**

**Example Trade Space Analysis: Quantifying HMD Contributions to TOC**

Added HMD mass is likely to increase TOC as shown previously in Figure 2 (entitled "TOC in context of HMD causal loop diagram"). It is important to understand the relationship between added HMD mass and the potential increases in TOC. In the occupational health domain, costs come in the form of chronic neck injury to operators. In the safety domain, costs are in the form of increased acute injury. This section provides an example method of quantifying increases in TOC due to chronic and acute neck injury. The methods used to quantify acute and chronic costs could be generally applied to other aspects of HMD related TOC in a full HMD trade space analysis.

**Occupational Health**

The AF HSI Handbook defines the occupational health domain as: "The consideration of design features that minimize risk of injury, acute and/or chronic illness, or disability, and/or reduce job performance of personnel who operate, maintain, or support the system (AF HSI Office, 2009)." For the purposes of this analysis, the occupational health domain cost of added head supported mass that will be considered is chronic neck injury. There is little documentation or literature data on the impact that increased head supported mass might have on the chronic neck injury or its long term musculoskeletal effects on users. Coakwell et al. wrote an in depth review article on the neck injury of fighter pilots (Coakwell et al., 2004). Regarding chronic neck injury, they report findings that repetitive exposure to high G forces is linked to early cervical spine degeneration (Coakwell et al., 2004). They also noted that frequent minor acute injury to the cervical spine predisposes people for more significant neck injury from trauma due to the weakening of the soft tissue supporting the spinal column (Coakwell et al., 2004). The costs of these injuries are difficult to quantify. The unknown nature of the long term effects of

heavier HMDs is concerning. This could potentially be a cost to readiness if pilots are unable to fly because of chronic neck injury. It also presents an unknown long term health care cost to the government. Future study is warranted to understand these issues more fully so that this component of the trade study can be further understood and applied within the cost benefit analysis. A notional equation for the probability of chronic neck injury ($P_C$), which is formulated as a function of the exposure to high G forces (and resulting neck loading – $load_{neck}$) over time is shown below. Neck loading could be further described as a function of additional parameters if desired, to include HMD mass, HMD CG, and expected accelerative input.

$$P_C = \int_0^t f(load_{neck})dt$$

An overall cost equation for chronic neck injury incorporates the probability of chronic injury equation combined with the number of pilots in the population of interest ($n$), and the medical costs to treat the chronic injury ($C_{C\_med}$). It will be assumed for chronic injury that the pilot completes his flying career, thus there is no cost to train a replacement pilot.

$$C_C = P_C \bullet n \bullet C_{C\_med}$$

**Safety**

The AF HSI Handbook defines the safety domain of HSI as follows:

"The application of systems engineering and systems management in conducting hazard, safety and risk analysis in system design and development to ensure that all systems, subsystems, and their interfaces operate effectively, without sustaining failures or jeopardizing the safety and health of operators, maintainers, and the system mission (AF HSI Office, 2009)." For the purposes of this analysis, the safety domain costs of added head supported mass will include the potential for increased injury during crash (rotary and transport aircraft), and increased injury during ejection for fighter aircraft.

Increases in head supported mass has the potential to increase the risk of acute operator neck injury if the pilot is subjected to accelerative environments, especially highly accelerative environments that can occur during ejection. Studies performed with human subjects in accelerative environments have repeatedly demonstrated significant increases in neck loads when the subjects wear an HMD than without the HMD when exposed to the same input acceleration pulse (Buhrman and Perry, 1994; Perry, 1998; Doczy et al., 2004). Injury due to a heavier HMD with an off-axis center of gravity (CG) in this environment could range from low severity strains and muscle tears to high severity cervical spine fractures and ligament ruptures (Buhrman and Perry, 1994). Perhaps this finding appears intuitive as increasing the mass of the head would be expected to result in an increased force when the head is exposed to acceleration. However, the human body is a complex mechanical system including a series of linkages and soft tissue connections, which have the potential to dampen an input impulse. Thus these studies have added much needed understanding on the effects of helmet mass on human neck response.

Risk curves are the foundation of an injury criterion (Pellettiere, 2012). They provide a defined relationship between neck loading and probability of injury which can be used to compare various HMD system configurations or quantify the injury risk of a prototype system during qualification or acceptance testing. A criteria tied clearly to a defined risk function allows for the acceptance of higher risk in the context of an overall system performance and cost analysis.

To quantify the safety portion of this HSI analysis, an improved pilot-scale frontal impact (Gx) AIS 2+ risk curve (Figure 6) was developed with a mathematical form similar to the National Highway Safety Transportation Administration (NHTSA) neck injury criteria formulation called the Nij (equation below) (Eppinger et al., 2000). The NHTSA risk curve was inadequate for application to the aviation ejection environment due to its inability to predict the 5% risk of AIS 2+ neck injury desired by the Air Force escape system oversight office, and because it has never been validated with human subject data. The improved risk curve was constructed using existing human subject testing neck data (n=67, 6G / 2kg, 8G / 1.6kg, 8G / 2kg experimental configurations) combined with cadaver data from published research (n=6, 32-39G) (Parr et al., 2013).

$$Nij = \left|\frac{Fz}{Fzcrit}\right| + \left|\frac{My}{Mycrit}\right|$$

This risk curve was applied to two additional data sets of human subject testing neck loading data, comparing the predicted injury risk of a 10G/1.4kg HMD test with a 10G/0kg HMD test. A statistically significant difference was observed in Nij values between the tests. The mean Nij for the 0kg and 1.4kg tests were 0.108 and 0.164, which predicted a 0.81% and 1.01% risk of AIS2+ neck injury respectively. The risk curve provides the ability to ascertain the difference in risk presented by different HMD mass configurations. It could also be applied to data from different accelerative loading conditions. While very small, these differences in injury prediction due to HMD mass provide a basis to quantify injury risk. This approach can be applied to other data and boundary conditions from HMD systems to quantify increases in TOC based upon acute injury risk due to HMDs.

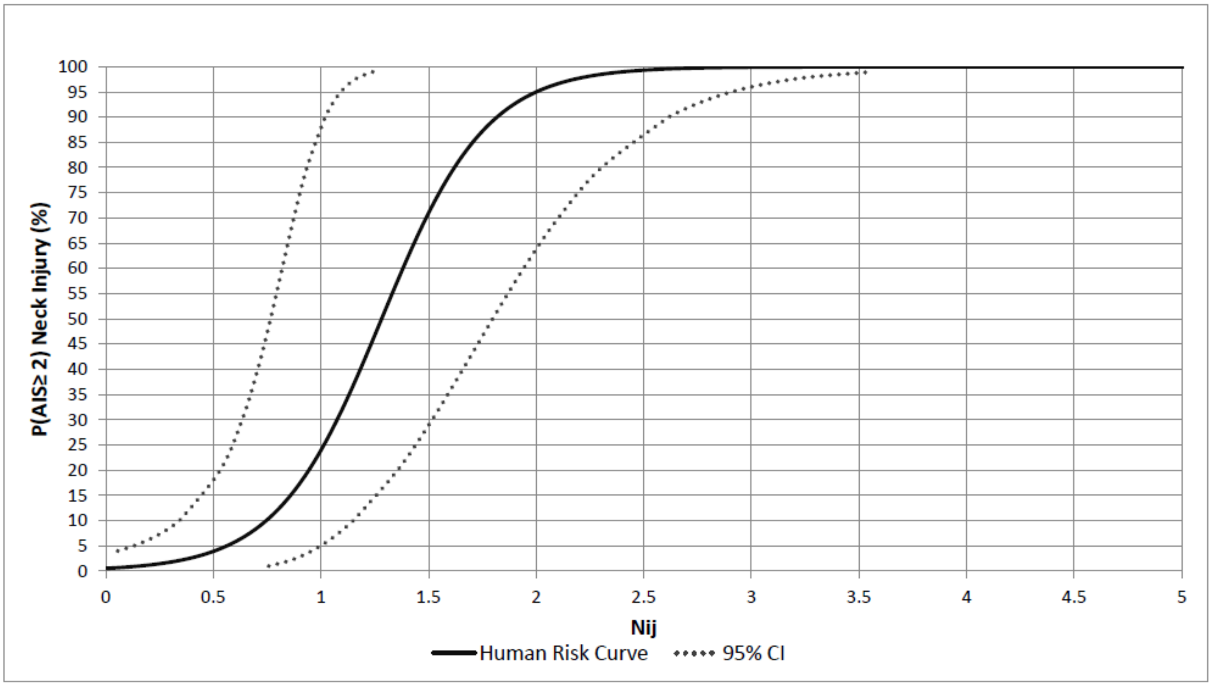


**Figure 6: Probability of AIS 2 or greater human Nij neck injury risk curves (95% CI show for human risk curve).**

Risk curves also enable estimates to be made concerning the cost of pilot neck injury from various HMD masses on life cycle costs using historical ejection rates, probability of injury (taken from the risk curve), and pilot replacement costs. While the example above uses the Nij as the loading input to the risk function, any desired neck injury criteria formulation could be used to determine the probability of acute injury ($P_A$). Similar to chronic neck injury probability, the probability of acute neck injury is a function of neck load ($load_{neck}$), and neck load can be treated as a function of HMD mass, HMD CG, and acceleration. Generically we can put this probability of acute injury into the form of the equation below.

$$P_A = f(load_{neck})$$

The probability of acute injury is incorporated into an overall cost function for acute injury ($C_A$) along with the number of pilots in the applicable population of interest ($n$), the medical costs associated with the acute injury ($C_{A\_med}$) , as well as the cost to train a replacement pilot ($C_{pilot}$) . The assumption will be made that an acute injury removes the pilot from any further flying duty.

$$C_A = P_A \bullet n \bullet C_{A\_med} + P_A \bullet n \bullet C_{pilot}$$

## Combined Performance and Cost Equations

Below are proposed equations that incorporate aggregated HSI performance and costs. Total system performance attributable to the HMD system can be quantified by the sum of each performance parameter.

$$p_{tot} = [p(x_1) + p(x_2) + p(x_n)] - [p(y_1) + p(y_2) + p(y_n)]$$

The cost equation is constructed based upon the definition of TOC. TOC includes the sum of HMD LCC ($LCC_{HMD}$), chronic injury costs ($C_C$), ejection injury costs ($C_A$) and other costs ($C_n$) which might be desired to include in the model. The major components of $LCC_{HMD}$ include research and development costs, investment costs, operating and support costs, and disposal costs (DAU, 2013a). TOC is minimized when each of the costs are minimized. This TOC equation contains only the portions of LCC applicable to the HMD system. In the portions of the equation where cost is tied to a probability ($C_C$ and $C_A$), TOC minimization occurs when the probabilities of chronic injury ($P_C$), acute injury ($P_A$), and other desired cost functions ($C_n$) are minimized.

$$TOC = LCC_{HMD} + C_C + C_A + C_n$$

Once fully described, the performance ($p_{tot}$) and TOC functions can be used to perform appropriate HSI optimizations. Considering the interpretation provided previously in the paper that HSI doctrine dictates maximizing performance and minimizing TOC, consider the equation below as a high level expression which aggregates the components of the trade study. The ideal HMD is one that maximizes the ratio of total performance ($p_{tot}$) to TOC over the parameters of $\overline{x}$ and $\overline{y}$. This expression might provide an appropriate method to consider the overall tradespace of performance benefits and costs associated with the added mass of an HMD.

$$HMD_{Ideal} = max_{\overline{x},\overline{y}}\left(\frac{p_{tot}}{TOC}\right)$$

## Summary

Applying HSI doctrine to a system optimization application like the HMD design trade space is a complex undertaking which spans a multitude of research communities. Quantifying the performance benefit and lifecycle cost elements required to perform a robust HSI trade analysis will necessitate targeted research. Future work will provide added understanding to each component of HMD performance and cost to more fully develop this model. In the short term, priority should be placed on creating first order approximations relating HMD system parameters to operator performance to demonstrate the extrapolation of this concept to a full model. In the longer term, targeted research should be conducted to specifically understand how common HMD system parameters such as night vision, FOV, resolution, and binocular systems influence operator performance to either validate the first order approximations or expand the functions to more robust representations.

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## Appendix B: Aviation, Space, and Environmental Medicine Conference Presentation

## Abstract – Assessment of the Applicability of the Nij Neck Injury Criteria to the Ejection Environment

**BACKGROUND:** Helmet Mounted Displays provide increased pilot capability but can also increase the risk of injury during ejection. A neck injury criteria appropriate for the aviation environment is needed to assess the safety of these systems and make design decisions during their development. **METHODS:** This study examined the applicability of the Nij criteria to evaluate helmets of varying weight. Data was analyzed from Air Force Research Laboratory human subject testing under various accelerative and head loading conditions to determine if the Nij construct was sensitive to changes in acceleration and helmet weight. It was also investigated if the Nij differed for male versus female subjects or was sensitive to variation in body weight, neck circumference, or sitting height due to variations in acceleration and helmet weight. **RESULTS:** A statistically significant difference in the Nij-specific neck loads of peak mean tension, compression, and flexion was observed when seat acceleration increased by 2 Gs with a constant 2 kg helmet. The 8 G, 2 kg test had significantly higher NTF and NCF than the 6 G, 2 kg test. No significant statistical differences in neck load were observed when helmet weight was varied from 1.6 kg to 2 kg at a seat acceleration of 8 Gs, though peak mean tension, compression, and flexion all increased. No injuries were observed despite the fact that the National Highway Traffic Safety Administration risk curves, which are based on peak observed Nij values, predicted a 5-7% risk of AIS3+ injury. Statistically significant differences were observed between "low" and "high" body weight individuals, but not between individuals having different gender, sitting height, or neck circumference. **DISCUSSION:** The Nij was sensitive to changes in acceleration and helmet weight, and somewhat sensitive to body weight. However, it would appear that the risk prediction of the criteria are not well developed for lower Nij values which are of most interest to military aviation. Military use of the Nij would require modification of the current criteria to better predict injury at lower force levels.

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## Vita

Lt Col Jeff Parr was born in Boulder, Colorado. After graduating from Fairview High School as valedictorian, he studied Civil Engineering at the United States Air Force Academy. He graduated with his Bachelor's Degree in Civil Engineering in May of 1998, earned the Outstanding Cadet in Physical Education award, and received Academic All-American honors for football. Following the academy, he was assigned to the 60th Civil Engineer Squadron, Travis AFB, California and worked as the Chief of Production Control and Logistics. He then earned a Master of Science in engineering management from the Air Force Institute of Technology, worked site activation for the F-22 program, deployed to Iraq in support of Operation Iraqi Freedom, and was a math instructor and football coach at the US Air Force Academy Preparatory School. After his assignment at the USAFA Prep School he was selected to attend Air Command and Staff College in residence at Maxwell Air Force Base. After Air Command and Staff College he served as the C-17 System Program Office Maintenance and Modifications Integrated Product Team Chief at the Aeronautical Systems Center, Wright-Patterson Air Force Base, Ohio. As Maintenance and Modification IPT Chief, he led the C-17 heavy maintenance and modification program for the C-17 fleet. He was then selected for an AFIT doctoral program in Systems Engineering. After earning his PhD he will serve on the AFIT faculty in the Department of Systems Engineering and Management. He has been married for 16 years to an amazing woman and has two fabulous kids. His parents and brother have been a tremendous support to him, and he is grateful to God for the blessings in his life.

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**14. ABSTRACT**

HMDs are becoming common human-machine interface equipment in manned military flight, but introducing this equipment into the overall aircraft escape system poses new and significant system design, development, and test concerns. Although HMDs add capabilities, which improve operator performance, the increased capability is often accompanied by increased head supported mass. The increased mass can amplify the risk of pilot neck injury during ejection when compared to lighter legacy helmets. Currently no adequate USAF neck injury criteria exist to effectively guide the requirements, design, and test of escape systems for pilots with HMDs. This research effort presents a novel method to develop neck injury criteria to aid the design and test of future HMD-centric escape systems. The pilot-scale injury criteria risk functions are constructed with combined human subject and post mortem human subject experimental data using a parametric survival analysis. The resulting neck injury criteria permit injury risk and classification levels specified by the Air Force escape system oversight office to be translated into system level test criteria. The application of the system level criteria during developmental and qualification testing of escape systems will ensure pilot safety and limit risk of neck injury.

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